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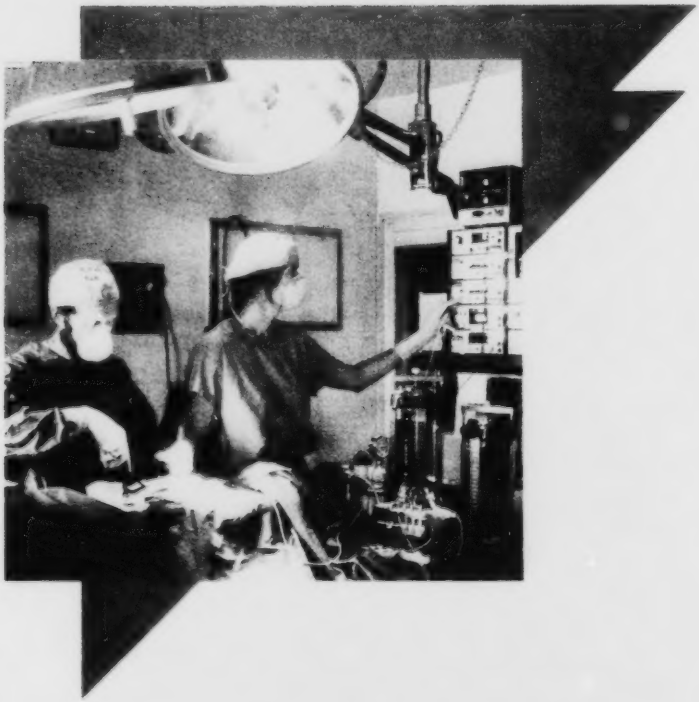
Biomechanical Assessment of Handrail Parameters

With Special Consideration to the Needs of Elderly Users

B.E. Maki and G.R. Fernie

August 1988



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89-03262/

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BIOMECHANICAL ASSESSMENT OF HANDRAIL PARAMETERS

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May 5, 1983

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Division of Building Research
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Contract No. OSX82-00180: Functional Capability of Elderly Persons Using Stair Handrails: The Influence of Handrail Height, Spacing and Section Geometry

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SUMMARY

The purpose of this study was to consider the effects of handrail parameters on stairway safety: (1) through analysis of the recommendations contained in the Occupational Safety and Health Administration (OSHA) Report of 1978, and (2) through experimental investigation. Special consideration was given to the needs of elderly users.

The OSHA recommendations were found to be deficient in many respects. Several recommendations regarding handrail shape, size and wall clearance were felt to be inappropriate, whereas the recommendations regarding handrail height were felt to be based on insufficient experimental data. The experimental study was aimed at correcting this insufficiency. Due to time constraints, handrail height was the only design parameter investigated, although there is no reason why the present experimental paradigm could not be used to study other handrail parameters.

In preventing a stairway fall, the functional role of the handrail is to generate stabilizing forces and moments. The horizontal and vertical forces counteract the undesirable forward and downward motion of the body. Of even greater importance is the stabilizing moment. Modelling the initial stages of a fall as inverted-pendulum rotation about the metatarsophalangeal joints of the trailing foot, the moment generated at the handrail counteracts the de-stabilizing moment due to body weight, as well as the rotational momentum of the body.

The experiment involved measurement of the maximum stabilizing forces and moments that subjects were able to generate, while standing stationary in an upright position. Changes in body position and orientation would be expected to change the forces and moments generated during an actual fall; however, static measurements of maximum voluntary effort (made in an erect posture) represent a reasonable approximation of the maximum stabilizing forces and moments that could be generated during the initial stage of a fall. It is felt that such measurements provide a suitable basis for the comparison of handrail heights.

The subjects were restrained so that handrail forces would be generated using the arm musculature alone. The handrail support was cantilevered and mounted to a force plate, so that all components of the applied force and moment could be measured. Measured forces and moments were non-dimensionalized with respect to each subject's destabilizing body weight and associated moment, in order to quantify the potential stabilizing effect relative to corresponding destabilizing factors. Non-dimensionalization allowed different subjects to be compared on a physically meaningful basis. Two age groups were tested: young (20 to 45) and elderly (59 and over), to examine possible age differences. Thirty-five subjects were tested.

Statistical analysis of the results showed that changes in handrail height did have a signficant effect on the ability to generate stabilizing forces and moments, and that significant age differences did exist in this ability. Moreover, all subjects showed a strong linear dependence on handrail height in generating stabilizing forces and moments. Ability to generate forward/backward forces and forward/backward moments increased linearly with increasing handrail height. Ability to generate upward force decreased linearly with increasing handrail height.

Based on the results, an optimal design range for handrail height was estimated to be 36 to 38 inches, inclusive. The mean subjective preference was found to be approximately 36 inches. These findings apply only to the test conditions (i.e. 1.75-inch diameter, cylindrical handrail, slope of 33 degrees) and cannot necessarily be extrapolated to other handrail shapes, sizes or slopes.

The results of the study should provide a useful basis for selection of handrail height in building design, where presently available strength data would not be applicable to handrail design. The present study is unique in that: (1) the test "handle" was positioned and oriented so as to simulate a stairway handrail, (2) stabilizing moments were measured, as well as forces, (3) subjects were allowed to choose their grip location and lateral displacement (a mock stairway was provided, to allow realistic choices to be made), (4) non-dimensional analysis was used to assess the efficacy of each handrail relative to each user, (5) both elderly and non-elderly populations were studied, and (6) a linear statistical analysis was performed to evaluate both age effects and the dependence of force/moment generation on handrail height.

RÉSUMÉ

Le but de la présente étude était d'évaluer les effets des paramètres relatifs à la rampe sur la sécurité des escaliers, premièrement par une analyse des recommandations contenues dans le rapport de 1978 de l'Occupational Safety and Health Administration (OSHA) et deuxièmement, par des expériences. On a tenu compte en particulier des besoins des utilisateurs âgés.

Les recommandations de l'OSHA se sont révélées insuffisantes à de nombreux points de vue. On a jugé en effet que plusieurs recommandations concernant la forme et la dimension de la rampe et sa distance par rapport au mur étaient inappropriées et que les recommandations concernant la hauteur de la rampe étaient basées sur des données expérimentales insuffisantes. La partie expérimentale visait à combler cette lacune. Le seul paramètre de conception que nous ayons eu le temps d'étudier a été la hauteur de la rampe. Cependant, rien n'empêche d'utiliser le présent modèle expérimental pour étudier d'autres paramètres relatifs aux rampes d'escalier.

Lorsqu'il s'agit de prévenir une chute dans un escalier, le rôle fonctionnel de la rampe consiste à produire des forces et des moments stabilisateurs. Les forces horizontale et verticale s'opposent au mouvement du corps vers l'avant et vers le bas que l'on cherche à empêcher. Le moment stabilisateur est encore plus important. Si les stades initiaux d'une chute sont modélisés sous forme d'une rotation pendulaire inversée dont le centre se situe aux articulations métatarsophalangiennes du pied arrière, le moment engendré à la rampe s'oppose au moment déstabilisateur dû au poids du corps ainsi qu'au mouvement de rotation du corps.

L'expérience consistait à mesurer les forces et les moments stabilisateurs maximaux pouvant être produits par des sujets stationnaires en position debout. On s'attend à ce que des changements de position et d'orientation du corps modifient les forces et les moments créés lors d'une chute réelle; toutefois, les mesures statiques de l'effort volontaire maximal (produit en position debout) constituent une approximation raisonnable des forces et des moments maximaux qui pourraient être engendrés au début d'une chute. On considère que de telles mesures peuvent servir de point de départ à la comparaison de la hauteur des rampes.

Les sujets étaient retenus grâce à des sangles, de sorte que les forces appliquées à la rampe ne provenaient que des muscles du bras. Le support de la rampe était placé en porte-à-faux et monté sur un plateau dynamométrique pour que toutes les composantes de la force et du moment appliqués puissent être mesurées. Les forces et les moments mesurés ont été transformés en valeurs adimensionnées par division, respectivement,

par le poids déstabilisateur du corps de chaque sujet et par le moment associé, ce qui a permis de quantifier l'effet stabilisateur potentiel par rapport aux facteurs déstabilisateurs correspondants. L'absence de dimension a permis d'effectuer une comparaison significative, basée sur des paramètres physiques, des différents sujets. Les sujets participant aux essais appartenaient à deux groupes d'âge : sujets jeunes (20 à 45 ans) et sujets âgés (59 ans et plus). On a ainsi pu examiner les différences possibles reliées à l'âge. Trente-cinq sujets ont participé à l'expérience.

L'analyse statistique des résultats a montré que la hauteur de la rampe influe beaucoup sur la capacité de produire des forces et des moments stabilisateurs et que cette capacité varie beaucoup en fonction de l'âge. De plus, chez tous les sujets, la production de forces et de moments stabilisateurs dépendait fortement et linéairement de la hauteur de la rampe. La capacité de produire des forces avant/arrière et des moments avant/arrière augmentait linéairement avec la hauteur de la rampe. Par ailleurs, la capacité de produire une force dirigée vers le haut diminuait linéairement avec la hauteur de la rampe.

D'après les résultats obtenus, la hauteur optimale de la rampe varierait de 36 pouces à 38 pouces inclusivement. La hauteur subjective préférée moyenne était d'environ 36 pouces. Ces constatations ne s'appliquent toutefois qu'aux conditions de l'essai (rampe cylindrique de 1,75 pouce de diamètre, pente de 33 degrés) et elles ne peuvent pas être nécessairement extrapolées au cas de rampes présentant des formes, des dimensions ou des pentes différentes.

Les résultats de la présente étude devraient être utiles pour choisir la hauteur de la rampe dans les cas où les données relatives à la résistance dont on dispose actuellement ne s'appliquent pas à la conception de rampes d'escalier. La présente étude est unique, pour les raisons suivantes : 1) la "poignée" utilisée pour l'essai a été positionnée et orientée de manière à simuler une rampe d'escalier, 2) les moments stabilisateurs ainsi que les forces ont été mesurés, 3) on a permis aux sujets de saisir la rampe où ils le voulaient et de choisir le déplacement latéral (un escalier factice a été installé pour leur permettre de faire un choix réaliste), 4) on a, dans l'analyse, utilisé des valeurs adimensionnées pour évaluer l'efficacité de chaque rampe pour chaque utilisateur, 5) on a étudié des populations âgées et jeunes et 6) une analyse statistique linéaire a été réalisée pour évaluer d'une part les effets de l'âge et d'autre part la relation de dépendance entre la production de force/moment et la hauteur de la rampe.

ACKNOWLEDGEMENTS

I would like to thank and acknowledge the following staff members at West Park Research for their invaluable assistance in completing this project: Richard Lobb, for constructing the experimental apparatus and for his help in running the experiments; Jean Holden, for designing and constructing the restraining harness, and for her help in running the experiments; Sheryl Bartlett, for her advice regarding the statistical analysis; Gerald Griggs, for his help in getting the electronics operational; and Gloria Redmond, for her help in recruiting volunteers to participate in the experiments.

I would also like to thank the experimental subjects for their participation.

Brian Maki
May 5, 1983

INTRODUCTION

Changes in handrail parameters (height, diameter, shape, wall clearance and horizontal spacing) might be expected to affect stairway gait in a number of ways. Possible effects might involve: (1) falling and loss of balance; (2) fatigue; (3) total energy expenditure; (4) rate of energy expenditure; (5) mean speed of ascent/descent for an individual user; and (6) mean traffic flow rate for multiple users. Although the effects of handrails on the energy requirements and speed of stairway gait may be significant, there is little doubt that their primary purpose lies in the prevention of falling accidents.

The purpose of this study is to investigate the effects of handrail parameters on stairway safety. The study is based on biomechanical and neurophysiological principles. Special consideration is given to the problems experienced by elderly stairway users.

The report is divided into five major sections. The first section is a general description of the nature of stairway falls and the role of handrails in preventing these falls. This discussion forms a basis for the sections which follow. The second section is a critical review of the recommendations contained in the Occupational Safety and Health Administration (OSHA) report [Armstrong, 1978]. The third section is a discussion of possible experimental approaches to handrail evaluation, whereas the fourth section is a detailed description of the experimental work which we performed in an attempt to evaluate the effects of changes in handrail height. The final section contains suggestions for further research.

BIOMECHANICS OF STAIRWAY FALLS

Introduction

With regard to stairway safety, handrails could conceivably serve two distinct functions: (1) prevention of loss of balance, and/or (2) prevention of falling once loss of balance has occurred. In the present context, "loss of balance" is defined as a condition characterized by abnormally large linear and/or angular acceleration of the body. A "fall" occurs when there is an impact between some part of the body (other than the feet) with the stairs. Thus, a fall is always preceded by a loss of balance; however, falling may be averted if appropriate recovery maneuvers are executed after the initial loss of balance.

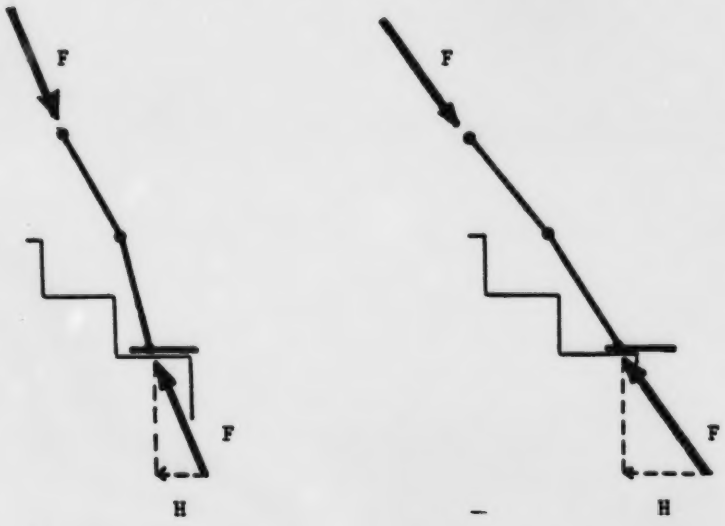
Prevention of Loss of Balance

In neurologically-normal users, loss of balance is apparently initiated either by slipping or by some form of misstep, that is, erroneous placement of the leading foot. A misstep could involve overstepping the nosing or catching the heel on the nosing in descent, or catching the toe on the riser in ascent [Alessi, 1979]. A lack of sufficient friction between the foot and the tread could result in slipping; however, this is unlikely since, under normal circumstances, the frictional forces in stair gait are very small (less than in level walking) [Harper, 1967]. If, during descent, the leading foot should overstride (landing with the heel near the nosing, or even missing a step completely), then the frictional force required to decelerate the body will increase and slipping is more likely. See Figure 1.1.

There are several reasons why missteps occur. Apparently, motor function during normal stairway gait is "pre-programmed" (i.e. control is effected by higher centres in the brain but without conscious effort); however, both formulation and execution of the program require visual input, or at least some prior knowledge of the stair dimensions [Craik, 1982; Freedman, 1976]. Thus, inadequate lighting or visual deficits in the user would be expected to contribute to misstepping. It has been suggested that the motor program is formulated while traversing the initial steps of the stairway, and that subsequent motor control is "automatic"; therefore, step-to-step variations in the stairway dimensions could render the motor program inappropiate and thereby cause a misstep [Templer, 1974]. It has also been shown that distractions (visual or aural) can lead to missteps [Templer, 1974], apparentally through a disruption of the execution of the motor program. Slips generally result from oversteps, as

(i) catching heel on nosing (ii) overstepping the nosing (iii) catching toe on riser

(a) types of missteps



(i) normal step (ii) overstep

(b) slipping: note that larger shear force (H) in overstep leads to increased probability of slipping.

Figure 1.1 - Causes of Loss of Balance on Stairways

discussed above, or in cases where the slipperiness of the stairway treads or user shoes is not recognized by the user (in creating the motor program).

Neurological disorders or deficits affecting the vestibular system, or motor control in general, may also cause loss of balance during stairway use. Investigation of the role of neurological dysfunction in stairway falls is beyond the scope of the present study.

Elderly stairway users are at greater risk than the general population. Reductions in hip, knee and ankle rotation during gait [Overstall, 1978] would be expected to correlate with an increased tendency for misstepping (e.g. catching the toe or heel). Missteps could also result from visual impairment, which is known to increase with age [Payton, 1983]. The elderly are also more likely to experience impaired postural balance, associated with a slowing of vestibular reflexes, impairment of proprioceptive feedback, and slowing of cerebellar processes [Overstall, 1978].

Handrail use may help to prevent loss of balance during stair gait. The exteroceptive feedback provided through continuous hand contact with the handrail may help to compensate for a lack of visual input and thereby prevent missteps or slips. Alternatively, generation of handrail reaction forces (see next section) during foot placement may stabilize the body and prevent loss of balance despite misstepping. Any significant horizontal force exerted by the handrail on the body will reduce the magnitude of shear force required at the foot-tread interface, and thus might act to prevent slipping. Use of the handrail to provide additional sensory information or to generate stabilizing forces is particularly important for elderly stairway users, whose sensory and motor deficits may lead to an increased tendency for misstepping.

Recovery from Loss of Balance (Prevention of Falling)

In general, a fall includes three significant components of motion: (1) rotation about the centre of gravity, (2) horizontal translation of the centre of gravity in the anterior-posterior direction and (3) vertical translation of the centre of gravity. Note that lateral translation is judged to be of secondary importance. The exact kinematics of the fall will depend largely on the nature of the misstep or slip that caused the initial loss of balance. In cases where the majority of weight-bearing occurs on only one foot and where there is relatively little knee and hip rotation in the weight-bearing leg, the falling motion can be approximated as pure rotation about a fixed axis. Thus, the human body can be modelled as an inverted pendulum, with rotation occurring about the supporting foot. This approximation would seem to apply to at least the initial stages of those falls resulting from missteps. In the case of slipping, weight-bearing may be distributed between the two feet, one of which is translating horizontally; thus, there is no fixed axis of rotation.

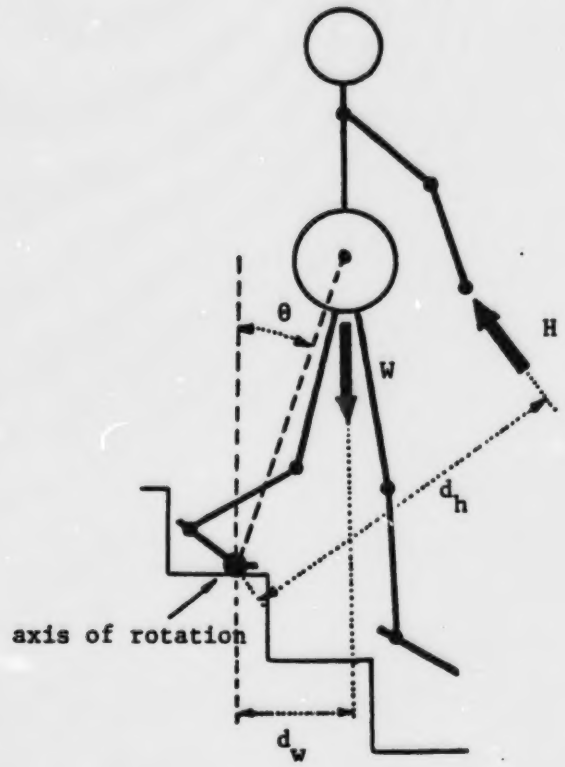
In fixed-axis rotation (misstep falls), the de-stabilizing moment (about the point of support) due to body weight is opposed by the stabilizing moment created by the ground reaction force. The ground reaction moment is transmitted from the foot to the body by means of the ankle musculature. In order to prevent a fall, the stabilizing moment must counteract both the de-stabilizing body weight moment and the rotational momentum of the body. The handrail serves to enhance postural stability and (thereby prevent falling) by increasing the total stabilizing moment about the foot. See Figure 1.2.

In the case of falls due to slipping and in the later stages of misstep falls (after substantial hip and knee rotation has occurred), the motion of the body can no longer be approximated as fixed-axis rotation. In order to analyze this type of motion, translation of the centre of gravity must also be considered. Normally, the vertical ground reaction force acts to limit downward acceleration, whereas the horizontal ground reaction force provides forward propulsion in ascent and limits forward acceleration in descent. If the supporting leg(s) "buckle", the vertical force transmitted to the trunk is no longer sufficient to counteract the downward acceleration due to gravity. In slipping, the horizontal component of the reaction force may be insufficient to propel the body forward (ascent) or to counteract the forward momentum of the body (descent). The handrail helps to prevent falling by generating additional stabilizing vertical and horizontal forces. See Figure 1.3.

Note that the force and moment applied to the body by the handrail are generated in reaction to the force exerted on the handrail by the hand. The magnitude and direction of the reaction force and moment are dependent on: (1) the orientation and position of the skeletal segments of the arm, and (2) the patterns of muscle activation occurring at the shoulder, elbow, wrist and hand.

In many cases, stairway users do not hold the handrail during normal ascent or descent. In the event of a loss of balance, such users must reach for the handrail and grab onto it. The time lag in this response allows the momentum of the falling body to increase and thus reduces the effectiveness of the handrail force and moment in stabilizing the body. For this reason, continous contact with the handrail is desirable.

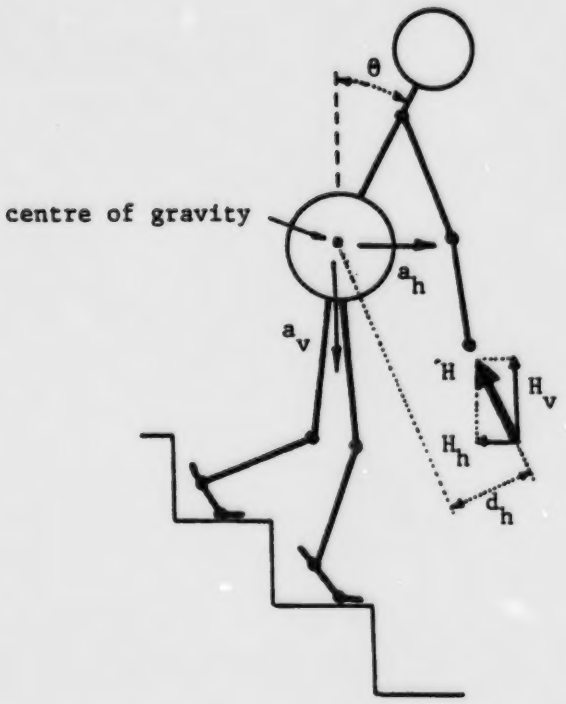
Elderly stairway users are less able to recover from initial loss of balance, due to a slowing of both postural reflexes and voluntary reaction time [Overstall, 1978] and due to a reduction in muscle strength [Payton, 1983]. In addition, the consequences of falling are more severe, due to an increased susceptability to bone fracture due primarily to osteoporosis [Brocklehurst, 1978].



Legend: θ = angle of inclination of inverted pendulum
W = body weight; d_w = moment arm of body weight
H = handrail force; d_h = moment arm of handrail force

Note: de-stabilizing torque = $W \times d_w$
stabilizing torque due to handrail = $H \times d_h$

Figure 1.2 - Fixed-Axis Rotation Model of Stairway Falls



Legend: $\ddot{\theta}$ = angular acceleration about centre of gravity
a_h = horizontal component of acceleration of centre of gravity
a_v = vertical component of acceleration of centre of gravity
H = handrail force; d_h = moment arm of handrail force
H_h = horizontal component of handrail force
H_v = vertical component of handrail force

Note: H_h opposes horizontal acceleration (a_h)
H_v opposes vertical acceleration (a_v)
stabilizing torque ($H \times d_h$) opposes angular acceleration about centre of gravity

Figure 1.3 - General Model of Stairway Falls

II REVIEW OF THE OSHA RECOMMENDATIONS

The following discussion will focus on the recommendations contained in the OSHA report [Armstrong, 1978]. In particular, the recommendations pertaining to handrail graspability, height, and wall clearance will be evaluated, on the basis of biomechanical and neurophysiological considerations, and on the basis of experimental results appearing in the literature. The discussion will also include special factors related to elderly stairway users.

Handrail Graspability

(1) OSHA Recommendations

"Handrails shall be circular, oval or oblong in cross-section so as to allow a power grip. Other forms of handrails (e.g. angle iron) shall not be allowed."

"Handrail circumference should be not less than 4.4 inches (11.2 cm) and no greater than 5.2 inches (13.2 cm). For cylindrical rails, this translates to a diameter of 1.4 inches (3.6 cm) and 1.65 inches (4.2 cm), respectively. The width (W) and thickness (T) for oblong handrails shall be determined using the following relationship:

$$4.4 \text{ inches} \leq (2W + 2T + 2\pi R) \leq 5.2 \text{ inches}$$

where $R \geq .25$ inch (1.24 cm)." (R is the radius of the corners)

(2) General Biomechanical Considerations

In selecting handrail parameters, the primary objective is to maximize the magnitude of the stabilizing reaction force and moment. As discussed in Section I, this force and moment is exerted on the body in reaction to the force exerted by the hand on the handrail. The force of the hand on the handrail is dependent on the joint positions and muscle activation in the arm.

The handrail force will have two components, one normal (perpendicular) and one tangential to the handrail surface. The relative importance of the two components will depend on the type of falling motion and on the size, position and orientation of the user's body. With regard to stabilizing moment, the tangential force component will tend to have a larger moment arm about the initial rotational axis (metatarsophalangeal joint). However, in general, it is not possible to predict which component is more important. For this reason, selection of handrail parameters should be aimed at maximization of both force components.

Note that the hand can also generate a couple (i.e. pure moment); however, the stabilizing effect of this couple with regard to rotation about the metatarsophalangeal joint is expected to be insignificant relative to the moment created by the handrail force.

(i) tangential force

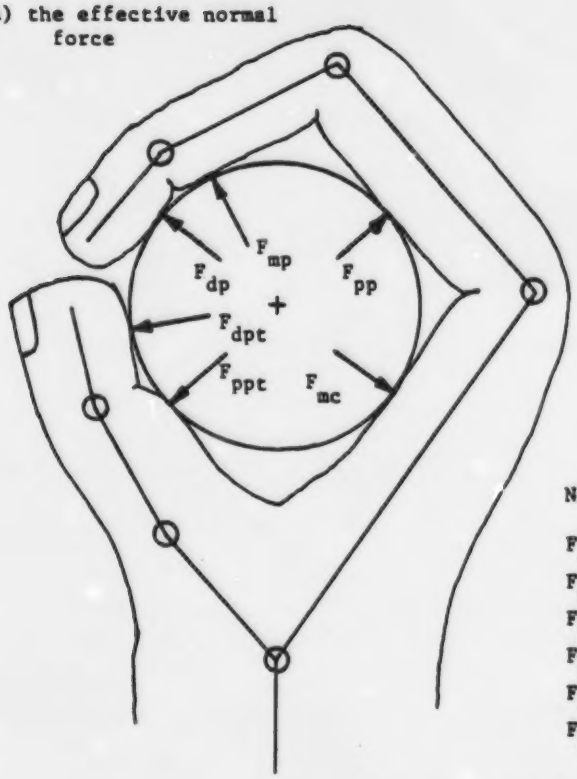
The tangential force component is generated through friction between the skin and the handrail surface. The magnitude of the maximum friction force that can be generated is dependent on the texture of the handrail material (as well as the skin) and the magnitude of the effective normal force.

Certain materials, when in contact with the skin, will allow greater frictional forces to be generated. In addition, "rough" (i.e. textured) surfaces will generally allow higher frictional forces than smooth surfaces. However, an aversive tactile response to these surfaces may obviate this advantage, in that users may reduce their gripping force. Unpleasant tactile sensation may also discourage continuous contact with the handrail during normal stairway use, which would tend to increase the danger of falling. If the hand should slip while grasping the handrail, a textured surface may tend to cause abrasive damage to the skin. Textured surfaces may also present hygienic problems. Research is clearly needed in this area, in order to clarify the trade-off between the benefits of increased coefficient of friction and the risks of aversive tactile response, abrasive tissue damage and hygienic problems.

As shown in Figure 2.1, the moments generated at the joints of the hand (through muscle activity) result in normal reaction forces at the points of contact between the hand segments and the handrail. Although each force will be distributed over a finite contact area, the effects can be approximated by a point load acting at the centre of pressure, as shown in the figure. The effective normal force, which determines the maximum frictional force, is the algebraic sum of these contact forces. For a given level of joint moment, the contact forces can be maximized by decreasing the moment arm between the centre of pressure and the proximal joint. Increasing the conformability between the hand and handrail will tend to increase the contact forces in this way, and may also increase the effective normal force by increasing the number of finger segments in contact with the handrail. Increasing contact area (through increased conformability) also has the advantage of allowing the contact forces to be distributed over larger areas, thus reducing the pressure on the soft tissue and the underlying nerves, arteries and veins. This will allow greater normal forces to be generated without incurring pain or tissue damage.

The contact forces can also be increased by increasing the muscle flexion moment at each joint of the hand. Joint moment is dependent on both muscle force (which depends on muscle length) and skeletal moment arm, and hence is dependent on joint angle. In general, maximal joint moment can be generated at some position lying between maximal flexion and maximal extension. Muscles which span more than one joint may impose additional constraints. For example, negligible flexion moment can be generated at the distal interphalangeal joints of the fingers unless there is significant flexion at the proximal finger joints [Tichauer, 1977].

(a) the effective normal force

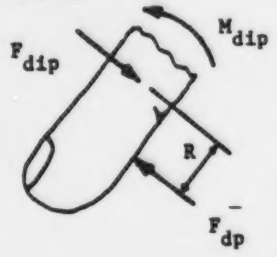


Normal Forces:

F_{dp} - distal phalanges
F_{mp} - middle phalanges
F_{pp} - proximal phalanges
F_{mc} - metacarpals
F_{dpt} - distal phalanx of thumb
F_{ppt} - proximal phalanx of thumb

$$\text{Effective Normal Force} = F_{dp} + F_{mp} + F_{pp} + F_{mc} + F_{dpt} + F_{ppt}$$

(b) typical free body diagram (distal phalanx)



F_{dip} = force at distal interphalangeal joint
F_{dp} = contact force from handrail
M_{dip} = muscle moment at distal interphalangeal joint

$$F_{dp} = F_{dip} = \frac{M_{dip}}{R}$$

Figure 2.1 Biomechanical Analysis of the Handgrip

To summarize, maximization of effective normal force will maximize the tangential force. This can be achieved by selecting a handrail size and shape that will: (1) minimize the moment arm between each contact centre of pressure and the proximal joint, (2) maximize contact area, so as to allow higher forces to be generated without pain or tissue damage, and (3) optimize joint angles so as to maximize the muscle moment generated at each joint of the hand.

This type of optimization problem cannot be solved directly without using experimental physiological data, because the functional redundancy of the muscles makes the system statically indeterminate [Chao, 1976]. However, knowledge of these factors can be used as general guidelines in selecting appropriate handrail shapes and sizes. In particular, the analysis indicates that the handrail design should allow each phalanx and metacarpal bone to contact the handrail in a position of moderate (e.g. midway between maximal flexion and extension) flexion. This requirement must be satisfied over a wide range of hand sizes. Norkin and Levange [1983] report that the position for optimal force generation in the finger flexors is 20 degrees of extension at the wrist, 45 degrees of flexion at the metacarpophalangeal joint, 30 degrees of flexion at the proximal interphalangeal joint, and 30 degrees of flexion at the distal interphalangeal joint (although no basis for this hypothesis is given).

While certain handrail shapes might maximize tangential force for certain hand sizes, it would seem that a circular shape would guarantee the best results over a wide range of hand sizes and shapes. A circular shape of sufficient diameter will always allow all hand, finger and thumb segments to contact the handrail in a flexed position. (In the limiting case, an infinite diameter would require full extension at all joints.) Non-circular shapes are less likely to satisfy this requirement for a wide range of hand sizes. In any case, a large radius-of-curvature is desirable, to avoid high contact pressures. A novel approach to reducing tissue pressure would be to select a handrail material with a suitably compliant surface layer, so as to allow maximum conformability between the hand and handrail surface.

For circular handrails, the minimum diameter can be selected to allow some upper limit of joint flexion (e.g. 60 degrees) in the largest hands (e.g. 95th percentile), whereas the maximum diameter can be selected to allow some lower limit of joint flexion (e.g. 20 degrees) in the smallest hands (e.g. 5th percentile). Maximum diameter might also be limited by the maximum "spread" angle between thumb and index finger, in the smallest hands. The required anthropometric data pertaining to hand and finger dimensions does not appear to be available at the present time, although estimates could be based on Garrett's data [1971].

An alternative approach is to compare different handrail shapes and sizes through experimental measurement of effective normal force or grip (i.e. squeezing) force. Apparently, no such experimentation has been reported in the literature. The studies which do appear in the literature (and which were cited in the OSHA report [Armstrong, 1978]) are generally irrelevant to stairway safety.

Ayoub [1971] recorded electromyographic acitivity (EMG) in the forearm flexors and extensors of subjects who were instructed to repetitively lower and raise a cylindrical handle against a pulley-supported weight load. Various handle diameters and loads were tested. The maximum grip force (i.e. squeezing force) exerted on the handle was measured during each trial; however, no measurements were made of maximal grip force per se (as erroneously stated in the OSHA report). Ayoub reported an optimal diameter based on maximal force/EMG ratio. While this parameter might be indicative of reduced tendency to fatigue during repetitive gripping tasks, it is not particularily relevant to handrail useage.

A number of investigators [Kraft, 1972; Swanson, 1970; Taylor, 1954] have studied grip force. In Swanson's study, power grip force was measured using a handle-based dynamometer, whereas pinch grip force was measured using an thin, instrumented disc. The specifics of the measuring apparatus used in the other two studies are not detailed; however, it is apparent that separate devices were used for testing power grip and pinch grip. These investigations were not intended to compare the two types of grip. Furthermore, the devices used to measure pinch grip strength did not simulate realistic handrail dimensions or gripping configurations (the tests required a prehensile, "three-jaw-chuck" type of grip, whereas a handrail grip would use all five digits). Nonetheless, the OSHA report cites these studies as evidence that the power grip can generate higher forces than the pinch grip and that handrails should be designed for power grip useage.

The power grip does allow for flexed contact of all hand segments with the handrail, whereas the pinch grip typically does not. Thus, the power grip satisfies the force optimization requirements described earlier and is indeed more likely to provide higher gripping forces than the pinch grip during handrail use; however, there is no experimental evidence to support this supposition.

Montoye and Faulkner [1964] found that grip strength varied only slightly as Stoelting-dynamometer handle-spacing was varied over a fairly wide range (corresponding to a circumference range of 5.0 to 5.8 inches), for a large range of hand sizes. Subjects with small or large hands experienced reduced strength at handle spacings outside this range.

This result is applicable to design of handrails having the same shape as the Stoelting dynamometer (i.e. oblong, with 3/4 inch diameter ends); however, the same circumference limits cannot be applied "carte blanche" to other handrail shapes (as is done in the OSHA report), since the biomechanics of the grip will obviously change. Also, note that the circumference range inferred in the OSHA report is incorrect, as the authors failed to account for the difference between Stoelting dynamometer scale readings and actual handle spacing. It should also be noted that dynamometer-measured grip force is not equivalent to effective normal force, since the

dynamometer measures only the opposing squeezing forces applied to the handles and does not include the lateral forces that would occur in gripping a handrail. Thus, the results are actually more applicable to the generation of normal pulling forces (see below) than to the generation of tangential forces.

Hertzberg [1955] also tested the effect of dynamometer handle-spacing on grip strength and found a single optimal spacing. However, this result is of limited value to handrail design, since (1) the subject population (Air Force pilots and cadets) was not representative of the general stairway user population, and (2) there were no attempts to analyze the effects of variation in hand size.

(ii) normal force

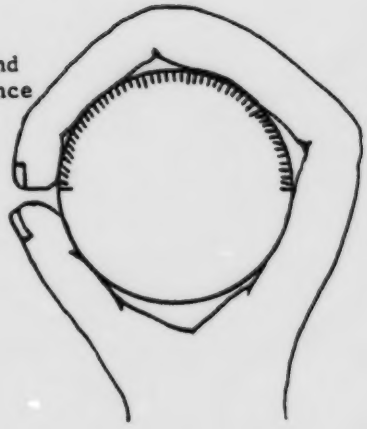
The biomechanical requirements for exerting a normal force on the handrail depend on the direction of the force. If the force is to be exerted away from the shoulder (i.e. a pushing action), the force is applied through the surface of the palm (ie. the thenar "heel" of the thumb) and requires little or no gripping force in the fingers. This situation requires a handrail with a large surface area of relatively large radius-of-curvature, allowing the force to be distributed over a sizeable tissue area, so that a large force can be exerted without discomfort or tissue damage.

If the force is to be exerted towards the shoulder (i.e. a pulling action), the force is applied primarily through the finger surfaces. The actual pulling force is created by the arm musculature, whereas the gripping muscles serve only to maintain the hand in a hook-like shape. The functional requirements for this task are similar to maximization of effective normal force, as discussed above, with the exception that our concern is now limited to those contact forces acting on the distal surface of the handrail. As mentioned earlier, dynamometer-based experimental data on grip force is applicable to this situation.

In generating tangential forces, the orientation of the hand relative to the handrail was not important. Now, however, the normal force exerted through the finger surfaces must be directed towards the shoulder, which requires contact on the distal surface of the handrail. In the limiting case, the handrail may be so large or of such a shape that there can be no hand contact on the distal surface (see Figure 2.2). The optimal handrail size will allow maximal contact area over the distal surface, to maximize the number of finger joints contributing to the total "pulling" force, and to minimize contact pressure. Contact pressure is very important because the arteries, veins and nerves in the fingers are easily compressed [Tichauer, 1977]. Maximal distal contact area requires a maximum diameter small enough to allow "wrap-around" for the smallest hands (e.g. 5th percentile) and a minimum diameter large enough so that there is no "interference" between the fingertips and palm for the largest hands (e.g. 95th percentile).

(a) no wrap-around

(b) maximum wrap-around
with no interference



(c) interference
(thumb not shown)

Figure 2.2 Wrap-Around and Interference in the Handgrip
(N.B. hatching indicates distal surface of handrail)

The line of action of the push/pull force should coincide with the natural thrust axis of the forearm, i.e. a line passing from the lateral edge of the elbow to the tip of the extended index finger [Tichauer, 1977]. Handrail shapes and sizes that require wrist extension or flexion (away from the optimal position) to achieve a comfortable (i.e. conformable) gripping position will shift this line of action, resulting in a bending moment about the wrist. This moment must be resisted by the wrist musculature, which may limit the amount of handrail force that can be generated. A handrail with a wide, flat upper surface would seem to create this sort of problem (e.g. oblong with major axis horizontal) and should therefore be avoided.

(3) Special Considerations for Elderly Users

Elderly stairway users face three major problems with regard to handrail "graspability": (1) loss of hand strength, (2) reduction in range of joint rotation, and (3) reduced elasticity in the skin and subcutaneous tissue [Payton, 1983; Roberts, 1960; Ma, 1950]. The loss of grip strength could conceivably be compensated to some extent through optimal design of the handrail shape and size, so as to maximize the normal and tangential forces generated by the available muscle moments. In addition, maximal generation of tangential forces could be achieved through the use of handrail materials with high coefficients of friction. The reduction in flexibility of the fingers may place limitations on the acceptable range of radius-of-curvature of the handrail surfaces. In particular, an expected tendency towards flexion of the fingers would limit the maximum allowable radius-of-curvature. Reduction in tissue elasticity will tend to concentrate the contact forces, resulting in higher contact pressures. This increased susceptability to pressure damage and/or pain requires large radius-of-curvature surfaces, to increase the load-bearing area and reduce the contact pressure. A compliant surface layer would help to increase conformability between the hand and handrail, and would thereby decrease tissue pressure as well.

Elderly users also experience losses in sensory function [Payton, 1983; Kenshalo, 1979]. In particular, there may be reductions in visual, tactile and kinesthetic capabilities. Tactile/kinesthetic sensing of body position relative to the handrail may contribute to postural stability during stairway use (particularily in users with visual deficits); therefore, handrail materials should be chosen to enhance the tactile sense. For example, since tactile responses require changes in skin deformation [Deatherage, 1972], tactile adaptation to skin contact with uniform surfaces might be avoided by using textured surfaces on handrails.

(4) Evaluation of the OSHA Recommendations

The shape recommendations are generally in agreement with the principle of using relatively large radius-of-curvature surfaces, so as to minimize contact pressure and allow power grip useage (i.e. flexed contact with all segments of the hand) over the wide range of hand sizes and shapes that exists. However, radially-assymmetric shapes, such as the oblong or oval, are unlikely to allow flexed contact or to provide adequate conformability for certain hand sizes and shapes; therefore, circular cross-sections are preferred. (Recall that flexed contact and conformability are important for maximizing both normal and tangential forces while generating minimal tissue pressure.)

If oval or oblong shapes are to be permitted, then it should be further stipulated that they must be orientated with the major axis vertical, since this orientation would tend to allow better conformability between the hand and handrail to be achieved without causing bending at the wrist (which would tend to reduce grip strength). In this orientation, the larger radii-of-curvature at the "ends" of the oblong shape tend to provide for greater distribution of normal forces (in pushing or pulling actions); therefore, the oblong is preferable to the oval (i.e. elliptical) shape.

The dimensional recommendations are highly questionable. From a biomechanical standpoint, specification of handrail circumference without reference to handrail shape is meaningless. Selection of the correct circumference will be of benefit only if the handrail shape conforms with the anatomy of the hand. Separate size recommendations are needed for each permissable handrail shape, based either on: (1) experimental measurements of grip force, or (2) anthropomophic analysis. The analytical approach would have to consider three requirements: (1) achievement of the flexed contact grip (i.e power grip), as required for maximal tangential and pulling forces, (2) maximization of "wrap-around" and minimization of "interference", as required for maxmimal pulling force, and (3) maximization of proximal surface area and radius-of-curvature, as required for maximal pushing force. Note that the analytical approach could be tailored to elderly user populations through the use of suitable anthropometric data (not presently available), whereas the experimental approach could use an elderly subject population.

The OSHA report also accepts the premise that handrails should have a smooth surface. This recommendation neglects the possibility of using textured surfaces to increase frictional force or to provide enhanced tactile feedback for elderly users. As discussed earlier, experimental work is needed to assess the trade-off between the benefits and problems of using textured, high coefficient of friction materials.

(5) Summary

The present analysis indicates that circular handrails are preferable to other shapes, including oblong or oval. The OSHA stipulation that the handrail shape should allow a power grip is valid, but is of no practical use to a building designer. Power grip useage should be guaranteed through specification of the handrail shape and size limitations. The dimensional limits in the OSHA report are based on an incorrect interpretation of experimental data and are therefore suspect. A more rigorous investigation of the problem (using anthropometric analysis and/or experimentation) would provide a more solid foundation for handrail dimension standards and is recommended. (Since appropriate anthropometric data is not presently available, the experimental approach might be more expedient.) The question of surface texture and compliance requires further experimental evaluation.

Handrail Height

(1) OSHA Recommendations

"Handrail height for all stairs, including ship's ladders, measured from the top surface of the handrail, to the tread surface at the leading edge of the tread, shall range from 31 inches (78.75 cm) to 33 inches (83.83 cm), with 33 inches the preferred value whenever possible."

(2) General Biomechanical Considerations

Absolute minimum and maximum limits on handrail height should be based on the ability of the user population to grasp the handrail without compromising postural stability. In particular, the handrail must not be so high or so low as to force the user to lean forward or sideways in order to grasp the handrail, since this would increase the de-stabilizing moment created by the body weight (see Figure 1.2). In addition, the user must not be forced to grip the handrail behind the body (in descent), since this requires the stabilizing force to be exerted primarily through a pulling action. As discussed earlier, pulling requires activity in the finger flexors and is expected to generate lower handrail reaction forces than pushing actions.

Within these absolute limits, the acceptable range of heights can probably be further refined through biomechanical considerations of the generation of stabilizing force and moment. As stated earlier, the primary goal is to maximize the stabilizing force and moment exerted by the handrail on the body. The stabilizing force and moment are generated in reaction to the force exerted by the hand on

the handrail. This force is generated by the musculature of the arm and shoulder, and is therefore dependent on the shoulder, elbow and wrist angles [Pauwels, 1980; Hazelton, 1975]. In addition, the location of the grip relative to the initial axis of rotation during a fall will influence the magnitude of the stabilizing moment.

Theoretically, it should be possible to create a biomechanical model of the arm and, through computer simulation, determine the optimal height for a given range of user heights (e.g. see Garg [1975]). However, present physiological knowledge cannot predict a priori the pattern of muscle activation because of functional redundancies between different muscles; therefore, experimental data would be required.

Furthermore, grip location is not constrained by handrail height, since the arm has redundant degrees of freedom in handrail useage. Therefore, the user's choice of grip location (during continuous handrail use) is determined by psychological factors, comfort being the most obvious consideration. For example, the user could attempt to: (1) minimize radial or ulnar deviation of the wrist (continual use of the arm in these positions is related to epicondylitis and tenosynovitis respectively [Tichauer, 1977], and might therefore be expected to cause discomfort), or (2) maintain extension of the shoulder and elbow, since this position would require minimal anti-gravity muscle activity. Since the comfort-optimizing strategies of stairway users are not known, and would be expected to show large user-to-user variation in any case, an analytical solution to this problem is not feasible.

An alternative approach is to experimentally measure the amount of force and moment that users can generate at different arm positions (corresponding to different handrail heights). Force-measuring experiments have, in fact, been conducted [Hertzberg, 1972]; however, the results are not applicable to handrail use, for the following reasons: (1) the force was measured using a vertical or transverse horizontal handgrip; (2) the subjects had no choice in the grip location; (3) the actual heights of the handgrip are not known; (4) the magnitude of stabilizing moment created by each handgrip force is not known; (5) the subjects were not restrained, and the handgrip forces were therefore constrained by the reaction forces acting on the body; (6) the test population was not representative of the general stairway user population (young males were used).

(3) Special Considerations for Elderly Users

Reduction in muscle strength and range of joint rotation [Payton, 1983; Roberts, 1960] might be expected to affect the response of elderly users to various handrail heights. To minimize the effects of the loss in strength, efforts should be made to optimize the handrail height, with respect to maximal force generation. Limitations on joint rotation (due to arthritis and/or soft tissue

contractures) would be expected to favour flexion at the shoulder and elbow [Zorzitto, 1983]. This would imply that handrails designed for elderly users should be higher than those designed for the general population, in order to allow these users to reach the handrails. In addition, it is expected that range of arm abduction will be limited in elderly users; therefore, small lateral displacements from the handrail will be desirable.

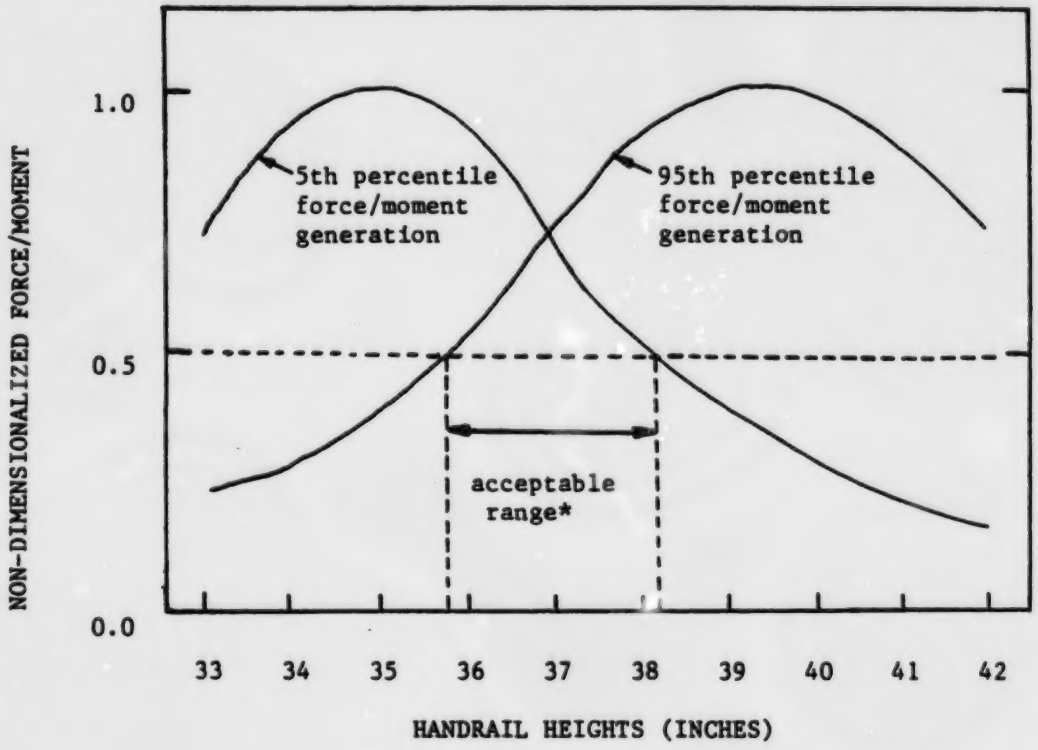
The elderly population is significantly smaller in stature than the general population [Stoudt, 1981; Roberts, 1960], which implies that handrail heights should be reduced on stairways with heavy useage by elderly people. Note that this recommendation for lower handrails conflicts with the earlier recommendation for higher handrails, which was based on range of motion arguments. Experimentation and/or anthopomorphic analysis is needed in order to assess the trade-off between the effects of reduction in stature and reduction in range of joint rotation in the upper limb.

(4) Evaluation of the OSHA Recommendation

The OSHA recommendation is based entirely on the ability of the user to grasp the handrail without adopting unusual body orientations (e.g. leaning forward or sideways) or inappropriate grasp locations (e.g. behind the body). The upper limit is based on an assumption that shoulder extension in excess of 90 degrees will be uncomfortable and fatiguing, during stairway ascent. The lower limit is based on the knuckle height of the 95th percentile male, with arms fully extended at the side of the body.

The recommendations do not take into account the amount of stabilizing force or moment that can be generated as the handrail height is changed. The lower limit could be too low, since full extension of the arm is not necessarily an optimal position for force generation. In addition, full arm extension will require ulnar or radial deviation at the wrist, which some users may find uncomfortable and therefore avoid (e.g. by not using the handrail, or by moving the grip location behind the body, either of which is undesirable).

Shorter users will be able to flex the elbow and/or shoulder or move the grip location forward. Either adjustment may improve comfort and will tend to increase the stabilizing effect of the handrail force during stairway descent, but only up to a point. As joint flexion is increased beyond some optimal point, the amount of handrail force that can be generated will start to decrease. The optimal force-generating positions, as determined for short (e.g. 5th percentile) and tall (e.g. 95th percentile) users, could be used to determine the acceptable range for handrail height, as illustrated in Figure 2.3.



* based on generation of ≥ 50% of maximum stabilizing force/moment for 5th and 95th percentile subjects

Figure 2.3 Determination of Acceptble Handrail Height from Force Data

The arguments for maintaining shoulder flexion below 90 degrees during ascent (i.e for comfort and to reduce fatigue) are reasonable; however, the analysis in the OSHA report is based on a questionable assumption that the elbow will be maintained in full extension during stairway ascent. In any case, safety during descent is felt to be a more important consideration; therefore, the maximum height should be determined using force generation considerations, as described above.

(5) Summary

The lower handrail height limit recommended in the OSHA report is acceptable as an absolute minimum value; however, the upper limit (based on stairway ascent considerations) is felt to be of questionable relevance. Appropriate force- and moment-generation experimental data are needed in order to provide a stronger biomechanical basis for handrail height recommendations. An experimental approach is of particular importance for elderly users. For these users, conclusions based on body dimensions may be inappropriate, due to the effects of other complicating factors, such as reductions in muscle strength and range of joint rotation.

Wall Clearance

(1) OSHA Recommendations

"Handrails built after January 1, 1979 shall have a minimum of two and one-fourth (2-1/4) inch (5.72 cm) finger clearance from any other object. If the physical surroundings do not prohibit greater finger clearance, then increasing the finger clearance above 2-1/4 inches may provide an added measure of protection during emergency or accident conditions."

(2) General Biomechanical Considerations

There are three major concerns in selecting handrail-wall clearance: (1) to allow for minimum response time in grabbing the rail during a fall; and (2) to minimize the danger of catching the arm between the handrail and the wall during a fall; (3) to allow sufficient space for grasping action to occur.

It is well known that positioning tasks of the upper extremity involve a trade-off between speed and accuracy [Fitts, 1954]. The more accurate a movement must be, the more time it takes. Grabbing onto a handrail during a fall could be modelled as a positioning task of this type. In this situation, potential impact of the hand with the wall might be perceived as a deterrent to an inaccurate

(overshoot) response. Thus, the handrail-wall clearance defines the minimum accuracy with which the grabbing motion must be made. Greater handrail-wall clearance reduces the accuracy requirements (as well as the perceived danger), and should therefore accelerate the grabbing response.

During a stairway fall, there is a chance that the user might catch his arm between the handrail and the wall. This could result in a number of injuries, including fracture of the wrist, dislocation of the shoulder, or, at the very least, bruises or cuts. The risk of this type of accident would seem to increase with the size of the handrail-wall clearance.

There is an obvious need to allow sufficient clearance for the fingers, and possibly the arm itself, so that an efficient grasping position can be achieved. However, the required clearance based on this criteria will be relatively small, and will likely be overshadowed by the clearance requirements for optimal reaction time.

(3) Special Considerations for the Elderly

Elderly stairway users will tend to have slower reaction times [Hasselkus, 1974], in grabbing for a handrail during a fall. Thus, larger handrail-wall clearance may be required in order to accelerate the grabbing response of these users. Experimental reaction-time data is needed to assess this possibility.

The increased susceptability of the elderly to bone fracture [Brocklehurst, 1978] must also be considered, in setting a maximum limit on handrail-wall clearance. Although a large wall clearance may increase the risk of "catching" the arm and sustaining a fracture, this type of occurrence may actually help to prevent a more serious fall from occurring. Therefore, limitations on maximum clearance are not felt to be warranted.

(4) Evaluation of the OSHA Recommendations

The OSHA recommendation is based on an estimate of the amount of clearance required to grasp the handrail from a vertical approach. As discussed in the report, the minimal clearance required to grasp the handrail from a horizontal approach (as would occur during a fall) would be significantly greater. (The actual approach which occurs during a fall is likely to be somewhere between the extremes of a vertical approach (best case) and a horizontal approach perpendicular to the wall (worst case).) The issue of accuracy/speed trade-off is not addressed. The problem of catching the arm between the handrail and wall is not considered, probably due to the fact that the small clearances recommended in the report would not allow this to occur.

Experimental evidence regarding the effect of perceived wall clearance on reaction time is needed. Such data would allow more relevant recommendations to be made with regard to minimum clearance. Assuming that this recommendation exceeds the minimum amount of space required for the hand to fit between the handrail and the wall, the possibility of injuring the arm by catching it between the wall and handrail becomes an issue. It would seem that enhanced prevention of falls outweighs the increased risk of arm injury, and so maximal clearances should be provided wherever possible. Obviously, building costs will be the major constraint, and so a minimum clearance recommendation should be specified, based on experimental reaction time data. Until such data is available, the recommendation should be based on the clearance required for horizontal grasping, i.e. 6.02 inches minus the diameter of the handrail (OSHA estimate). Note that this estimate is conservative, since it is derived by subtracting the 5th percentile thumb crotch height from the 95th percentile hand length. A more realistic estimate would be based on 95th percentile thumb crotch-fingertip length; however, such data is not available.

(5) Summary

For general stairway user populations, experimental evidence is needed to assess the effects of handrail-wall clearance on grabbing reaction time. It is not necessary to give special consideration to the slower reaction times of elderly users, since these users tend to maintain continuous contact with the handrail. Until such data is available, sufficient clearance should be provided so as to allow a grasping of the handrail from a fingers-extended, horizontal approach (i.e. the worst case), perpendicular to the wall, based on the 95th percentile hand. Thus, the minimum wall clearance should be approximately 6 inches minus the diameter of the handrail, and not 2.25 inches (as recommended in the OSHA report).

III METHODS FOR EXPERIMENTAL EVALUATION OF HANDRAIL PARAMETERS

As discussed in the first section, there are two possible mechanisms by which handrails could enhance stairway safety: (1) prevention of loss of balance, and (2) prevention of falling, once loss of balance has occurred (i.e. recovery from loss of balance). In this section, possible methods for experimental assessment of these two mechanisms are examined.

Prevention of Loss of Balance: Methods for Assessment

There are at least three approaches that could be followed in addressing the effects of changes in handrail parameters on the prevention of loss of balance on stairways: (1) observation of the incidence of missteps and/or slips, (2) measurement of the shearing forces on the foot, and (3) analysis of stair gait patterns.

(1) Observation of Incidence of Missteps and/or Slips

Misstep incidence could be measured through the use of foot switches, whereas measurement of slipping incidence would require some form of visual observation (i.e. human observer, film or video). Such experiments would require the experimental subject to make a large number of repeated traverses of each test stairway in order to obtain statistically significant data. The incidence rate could be accelerated by depriving the subject of certain sensory inputs (e.g. vision) or by providing visual or aural distractions, thereby simulating certain causal factors. Obviously, some form of safety harness would be required in these types of experiments.

Both foot switches and photogrammetric techniques (film or video) are commonly used in the study of level-surface gait [Chao, 1978]. Note that the foot switch technique has been used in at least one previous stairway experiment [Templer, 1974], although the significance of the protocol used is questionable. Templer used two switches, one supposedly relating to toe placement and the other to heel placement; however, both switches were located a fixed distance from the toe of the shoe for all subjects. It would seem that a more useful approach would be to relate the switch locations to the functional anatomy of the foot, that is, to place a switch at each potential weight-bearing surface (e.g. toes, metatarsal heads and heel). A variation from the predominant pattern (i.e. sequence of switch activation) of foot contact, as determined for each subject, would constitute a misstep.

(2) Measurement of Shearing Forces at the Foot

Possible problems with slipping could be evaluated by measuring anterior-posterior shear force at the foot-tread interface. This could be accomplished by installing a force plate at one of the stairway treads. Any handrail configuration which results in reduced shearing forces presumably decreases the probability of slipping. Repeated traverses of the test stairway would probably be required in order to obtain statistically significant results. The force plate approach is commonly used in the study of level walking, and has also been used in previous studies of stairway slipping [Harper, 1967; Esmay, 1964]. Harper also filmed the gait of his subjects while conducting his experiments.

(3) Analysis of Stair Gait Patterns

The effects of changes in handrail parameters on stair gait patterns could be studied and the results compared with normal stair gait patterns, which have been documented to some extent [Andriacchi, 1980; Townsend, 1978; Morrison, 1969; Joseph, 1967]. The underlying assumption is that certain abnormalities are indicative of a decrease in postural stability. Possibilities include abnormal patterns of muscle activation, joint rotation (hip, knee or ankle) or magnitude of the ground reaction force. Sensory deprivation or environmental distractions could be used to accentuate the abnormal patterns.

Appropriate measurement techniques are well-known: muscle activation can be recorded by means of non-invasive surface electrodes [Basmajian, 1978], joint rotation can be measured directly using goniometers or indirectly using stereometric methods (film, video or optoelectric detector) [Chao, 1978], and ground reaction force can be obtained using a force plate.

There are three potential problems with this appoach: (1) there is a lack of normative stair gait data for elderly subjects, (2) the normative data that is available is based on a relatively small sample size, and (3) it may prove difficult finding some simple, measurable gait parameter that will correlate directly with postural stability.

Recovery from Loss of Balance (Prevention of Falling): Methods for Assessment

There are three general ways to evaluate the effectiveness of any given handrail configuration in preventing falls: (1) incidence observation studies, (2) fall simulation, or (3) biomechanical testing.

(1) Observation of Incidence of Falls

The incidence of falls occurring on a public stairway could be recorded through direct observation or filming. One advantage of this approach is that the incidence of missteps/slips and the incidence of recoveries from missteps/slips could be studied, in addition to the incidence of falls.

The incidence observation approach does have a number of serious disadvantages. Uncontrolled variables, such as speed of travel and multiple-user traffic patterns could influence results in a random manner. Variation in the subject populations using each handrail configuration would also tend to randomize the results. These randomizing effects, in combination with the low incidence rate of falls, mean that extremely large numbers of users would have to be observed, in order to obtain statistically significant differences between the different handrail configurations.

In order to obtain the same subject population for each test handrail configuration, all the handrail configurations would have to be installed (serially) on the same stairway. However, the transitions from each handrail design to the next could be expected to cause a significant number of missteps and slips in themselves, and would therefore tend to invalidate the results. In addition, this approach would seriously limit the number of handrail configurations that could be tested.

An alternative approach would involve observing the fall incidence using a mock-up stairway in a laboratory environment. This would guarantee control over the subject population and would allow more handrail configurations to be tested. However, as in the "field" studies, statistically significant results would require either an extremely large number of subjects or a large number of trials for each subject.

(2) Simulation of Falls

Fall simulation involves perturbing the experimental subject's balance in such a way as to adequately simulate an actual fall. The effectiveness of a particular handrail is evaluated on the basis of the ability of the subject to recover from the initial loss of balance without sustaining a fall. The postural perturbation can be designed so as to simulate the effects of a misstep or slip during stair gait, and can be applied in the form of a disturbing moment/force or as a rotation/displacement of a supporting platform (representing the stair tread).

The most obvious problem in the fall simulation approach is the question of subject safety. Even if the risk of actually falling can be minimized through appropriate use of safety harnesses and placement of foam cushions, there remains a distinct risk of musculo-skeletal

injury occurring either as a direct result of the postural perturbation or as a result of restraining forces from the safety mechanisms. This risk is particularily serious in elderly subjects, who might be susceptible to bone fractures as a result of skeletal deficiencies (e.g. osteoporosis), muscle damage as a result of muscle rigidity, or ligament damage as a result of inadequate muscular stabilization of the joints.

Fall recovery involves a complex interaction of kinematic and neurophysiological factors. The motion of the body following a loss of balance is determined by the initial kinematic variables (linear and angular velocity and acceleration of each body segment), the mass and moment of inertia of the body segments, the moment and stiffness generated at each joint by muscle activity, and the reaction forces applied to the body through the feet and hands. The muscle activity which generates the reaction forces and the joint moments can occur involuntarily, in response to the motion of the body and limbs (excitation of vestibular and/or proprioceptive/exteroceptive reflexes), or through voluntary action. Failure to simulate accurately the interaction of all these factors could lead to questionable results.

Accurate simulation of a fall is a difficult problem. The forward momentum of the body which is present during stair gait undoubtedly contributes to the dynamics of a fall; therefore, perturbations applied to a static subject will not be an accurate representation of a stairway fall. It would seem that the most reliable way to duplicate the initial kinematic conditions is to provoke a stumble or slip as the subject traverses a test stairway. However, in this type of experimental situation, subject apprehension, as well as learning effects, might be expected to influence the results. Furthermore, the safety concerns mentioned earlier are obviously even more critical in this type of experiment.

(3) Biomechanical Tests

The alternative to fall simulation is to evaluate the performance of each handrail configuration by measuring pertinent biomechanical variables. Obviously, the variables selected must somehow relate to postural balance during actual stair gait. The absolute measurements are used as a means of comparison of the different handrail configurations only as they affect the performance of each individual subject. The measurements can be used to compare different subjects only if they are appropriately non-dimensionalized, since individual characteristics (e.g. body weight, height, somatotype, etc.) unrelated to handrail parameters will affect absolute performance.

One possibility is to study postural sway in subjects as they stand on an sinusoidally-oscillating platform while holding onto a stationary handrail. This approach has been used to study postural balance in

free-standing subjects [Holliday, 1983], and has been demonstrated to be a safe and acceptable test for use with elderly subjects. Increased postural sway implies a reduced ability to counteract the de-stabilizing moment associated with the body weight and is therefore expected to indicate an increased tendency to fall under the influence of more abrupt perturbations. Such a correlation between postural sway and observed falling behaviour has in fact been demonstrated in similar experiments where sway was measured in elderly subjects standing on a stationary platform [Fernie, 1982].

The major drawback to the postural sway approach in the current study is that the sensitivity of the test may not be sufficient to allow meaningful conclusions to be drawn. In other words, postural sway during the handrail-assisted experiments may be so small that changes in handrail parameters will not have a statistically significant effect. Attempts to increase the sensitivity through more abrupt platform oscillations may create safety problems.

An alternative biomechanical approach is to study the maximum forces and moments that the subject can exert on each handrail configuration. As discussed earlier, the stabilizing effect of the handrail is accomplished by increasing stabilizing moment and/or horizontal and vertical forces; therefore, increased moment/force is indicative of an enhanced ability to counteract de-stabilizing moments/forces and to thereby recover from loss of balance without falling.

The forces exerted on the test handrail could be measured by instrumenting the handrail with appropriate load cells. Stabilizing forces corresponding to the various falling situations could be obtained by asking the subject to both push and pull on the handrail in both horizontal and vertical directions, with as much force as possible. Stabilizing moments corresponding to the various falling situations could then be estimated by calculating the moment (due to the measured forces) about appropriate axes of rotation.

The maximal stabilizing forces/moments should reflect only the biomechanics of the arm and shoulder; therefore, the subject's body would have to be restrained during the testing, so that movement of the body could not influence the results.

One problem with this approach is that there is no way to ensure that the subject will always provide the same level of effort throughout the experiment. In addition, psychological and physiological fatique may cause the forces and moments exerted to decrease during the course of an experimental session. To compensate for fatigue effects, trials for the different handrail configurations would have to be repeated at least once, in reverse order. To ensure repeatable level of effort, electromyographic activity (EMG) could be monitored in the major muscle groups (e.g. biceps and triceps) and the measured forces/moments could be normalized with respect to the EMG recorded

for a pertinent muscle (e.g. biceps for pulling upwards, triceps for pushing downwards). However, normalization with respect to EMG is based on the assumption that the same pattern of muscle activation is used in performing a given task, independent of changes in the handrail parameters. Since this assumption is somewhat questionable, EMG normalization is not recommended. An alternative approach is to randomize the fatigue and motivational variation by testing the various handrail designs in a random order.

Note that the force/moment measurements described above correspond to the forces and moments that the subject would exert on the handrail while standing in an erect position. It seems likely that the values would change during stair gait, in which the hips, knees and ankles may be flexed to varying degrees, and during a fall, in which the angle of the trunk might also change markedly. However, falls are expected to begin near the time at which the leading foot contacts the stair, and an erect posture is a fairly good approximation of body position and orientation (relative to the handrail) occurring at that instant (in both ascent and descent). Furthermore, there is no reason to suspect that the trends in relative handrail performance indicated by the stationary, erect measurements would not also hold during actual stair gait and during an actual fall.

Recommended Experimental Paradigm

In neurologically-normal stairway users (young and old), ability to recover from loss of balance is judged to be more important than the prevention of initial loss of balance in preventing serious stairway accidents. Therefore, given the limited time and resources allotted to this study, the present experiments will address the issue of fall-prevention (i.e. recovery after loss of balance). It is recommended that the role of handrails in preventing loss of balance be further investigated at a later date, using the misstep/slip incidence observation, shearing force measurement and/or gait analysis methods described earlier.

Falling-incidence observation studies are not feasible for logistical reasons. Time constraints rule out any possibility of observing the large number of users or trials that would be necessary to obtain statistically significant results.

The fall simulation techniques: (1) are unsafe, particularily for use with elderly subjects, (2) are too difficult to implement in a manner that accurately simulates actual stairway falls, and (3) involve so many uncontrolled variables that useful results will not be obtained without conducting excessively large numbers of experimental trials.

The biomechanical testing approach appears to be the most appropriate choice for this study. The force/moment measurement approach is recommended in preference to the postural sway method. Using safe

levels of platform oscillation, sway measurement is likely to be too insensitive to identify the optimal handrail configuration. On the other hand, force/moment measurements are expected to vary markedly with changes in handrail parameters, for the following reasons: (1) changes in shoulder, elbow and wrist angle will affect joint moments (and hence handrail reaction force and moment) as a result of changes in muscle lengths and skeletal moment arms, (2) changes in the preferred location of the grasp relative to the relevant axes of rotation will affect net stabilizing moment, and (3) changes in the mode of grasping will cause changes in the grip pressure which in turn will affect the handrail reaction force and moment. In addition to the biomechanical measurements, the subjective preferences of the subjects could be recorded.

The force/moment measurement approach has a number of limitations which should be fully understood: (1) although voluntary exertion of maximum force is a commonly accepted measurement in anthropometric studies [Roebuck, 1975], variation in level of effort and fatigue effects may influence the results to some extent; (2) the erect posture of the body and the location of the grasping hand on the handrail used in the tests are only an approximation of the actual conditions at the beginning of a fall; (3) the measured values of moment and force approximate the maximum values that could be exerted at the beginning of a fall, but there is no guarantee that these maximum values would actually occur; (4) the maximum values of moment and force would be expected to change as body position and orientation change during the course of a fall; (5) the static values of moment and force measured in the tests would be expected to differ from the dynamic forces and moments exerted during an actual fall due to such effects as changes in muscle contraction velocity and changes in tissue perfusion.

In spite of these limitations, the static force/moment measurements are felt to provide a valid means of comparing the effects of changes in handrail parameters. Although the stabilizing force and moment exerted during an actual fall may vary considerably due to the dynamic and positional effects discussed above, increased capability for exertion of stabilizing force and moment (measured statically in an erect position) would be expected to correlate with an increased probability of recovery from loss of balance during a fall.

IV EXPERIMENTAL EVALUATION OF HANDRAIL HEIGHT

The number of handrail parameters that could be evaluated was severely limited by time constraints. We elected to concentrate our efforts on the study of handrail height. Handrail shape, diameter and horizontal spacing (between parallel handrails) were not studied, although there is no reason why the present experimental paradigm could not be used to evaluate these parameters. Spacing from the wall is a critical factor in those falls in which the user must grab for the handrail, and could probably best be tested (at some later date) by means of simple simulation experiments (e.g. measure grabbing response time and incidence of "misgrabs" for different wall spacings).

As discussed in the previous section, the most practical way to study the effects of changes in handrail height is to measure the generation of stabilizing forces and moments under static conditions of maximal voluntary effort. This is the approach that we used in our experimental investigation.

Methodology

Since descent falls tend to be much more serious than falls occurring during ascent, the testing was limited to the descent condition. Handrail cross-sectional shape, diameter, surface texture, and slope were held constant during the experiments. A circular cross-section was selected, since this is felt to be the optimal shape, as discussed in Section II. A diameter of 1.75 inches lies in the middle of the optimal circumference range estimated from the data of Montoye and Faulkner [1964], and was therefore selected for use in the experiments. The handrail was constructed of oak; the surface was sanded and coated with matte-finish polyurethane (Flecto Varathane II, Super 101). A slope of 33 degrees was chosen, since this value lies midway within the range of preferred slopes (i.e. 30 to 35 degrees) indicated in a number of standards [Nelson, 1973]. Note that this slope is also the maximum recommended by Pauls [1982].

Six handrail heights were tested within the range of 32 to 42 inches. (Handrail height was measured from the leading edge of the stair tread, as is the common practice.) The order of testing was randomized, so as to negate learning and fatigue effects. The initial height was used as a learning trial to allow the subjects to gain familiarity with the protocol, and was re-tested at the end of the session. Thus, each subject underwent a total of 7 trials. Allowing 5 minutes for each trial (including rest periods, to minimize subject fatigue) plus 20 minutes set-up time, the total testing time for each subject was generally 50 to 55 minutes.

The subject was allowed to select his/her preferred lateral displacement from the handrail by imagining that he/she was using the handrail while descending a stairway. Similarily, at each handrail height, the subject was allowed to select a grip position on the handrail. To aid the subject in choosing lateral displacement and grip location, the tests were conducted on a mock stairway.

All heights were tested at the lateral displacement chosen by the subject. Originally, we had hoped to repeat the tests at maximal lateral displacement; however, it became evident that this would make the duration of the experimental session unacceptably long for the subjects. Furthermore, conditions of maximal lateral displacement are likely to occur when the stairway user is forced to grab for the handrail. In these situations, force/moment generation is probably less important the the speed and accuracy of the grabbing response. (Note: distributions of the preferred lateral displacements are plotted in Appendix B; this information may be of use in determining optimal displacement between parallel handrails.)

In order to minimize motivational variation, all subjects were given a standard set of instructions. In addition, spectators were not allowed, and no verbal encouragements were given by the experimentor. At each handrail height, the subject was given a sequence of four instructions: (1) "push forward along the handrail", (2) "pull backwards along the handrail", (3) "push downwards towards the ground", and (4) "pull upwards away from the ground". These directions were found to be readily understandable to the experimental subjects, whereas instructions to generate maximal forces and moments in given directions would undoubtedly confuse many subjects. Subsequent data analysis examined the maximal forces and moments generated at each handrail height, regardless of which of the four instructions happened to result in those maximal values.

During the tests, the subjects were restrained by means of straps and a vertical back-board. The restraining straps passed across the pelvis, across the sternum and over the shoulders (see Figure 4.1). The restraint had two effects: (1) the subjects were forced to generate the forces and moments using only their arm musculature, not their body weight, and (2) the forces and moments generated were not limited by inadequate reaction forces acting on the body. To minimize discomfort and fatigue, subjects were allowed to move their feet between force measurements and were given the option of taking a rest (sitting) midway through the session. No subjects felt it necessary to take advantage of this option.

Prior to the experimental session, various body dimensions were measured, for use in later data analysis. Pertinent medical history was also recorded.

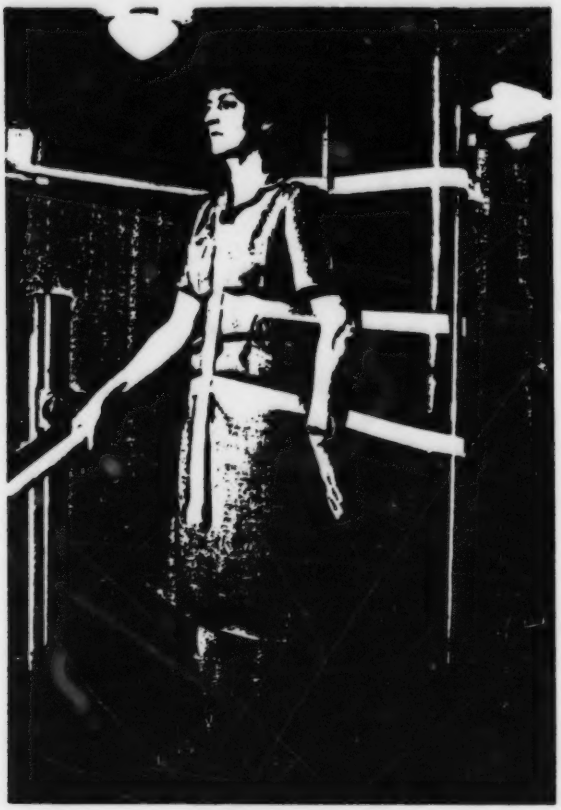


Figure 4.1 The Restraining Harness

Subject Selection

In order to minimize variability in the experimental results, the study used only normal subjects, that is subjects with: (1) no evidence of significant neurological, vestibular, arthritic or musculoskeletal disorders or deficits, (2) no history of abnormal problems with dizziness or falling, and (3) no present regimen of drugs or medications that might affect neuromuscular function. It is felt that studies of subjects with specific disorders should be conducted separately, to allow differences from the normal population to be identified.

In the time available, it was possible to recruit and test a total of thirty-five subjects. Eighteen subjects were elderly (59 or over), and the other seventeen subjects were selected to fall between the ages of 20 and 45. In order to allow extrapolation of the experimental results, subjects were selected at random within each age group. Thus, no attempts were made to recruit particular body sizes or ages, or a particular number of males or females. However, to simplify the protocol, all recruits were right-handed, and all tests were conducted with the handrail on the right-hand side.

The non-elderly subjects were recruited from the staff and students at West Park Hospital. The elderly subjects were recruited from visitors and volunteers at the hospital, and from relatives of hospital staff.

The characteristics (age, height and weight distribution) of the subject populations are plotted in Figures 4.2 to 4.4. Data for the general population is also shown, where available. Note that the height and weight distributions are approximately normal. In addition, the elderly subject population is somewhat smaller in stature and heavier in weight than the young group, on the average. These results would tend to indicate a subject population that is reasonably representative of the general population. However, note that the age distributions are both skewed in a manner which may or may not represent the general population. Also, note that the male/female ratio is somewhat higher in the subject population than would be expected in the general population.

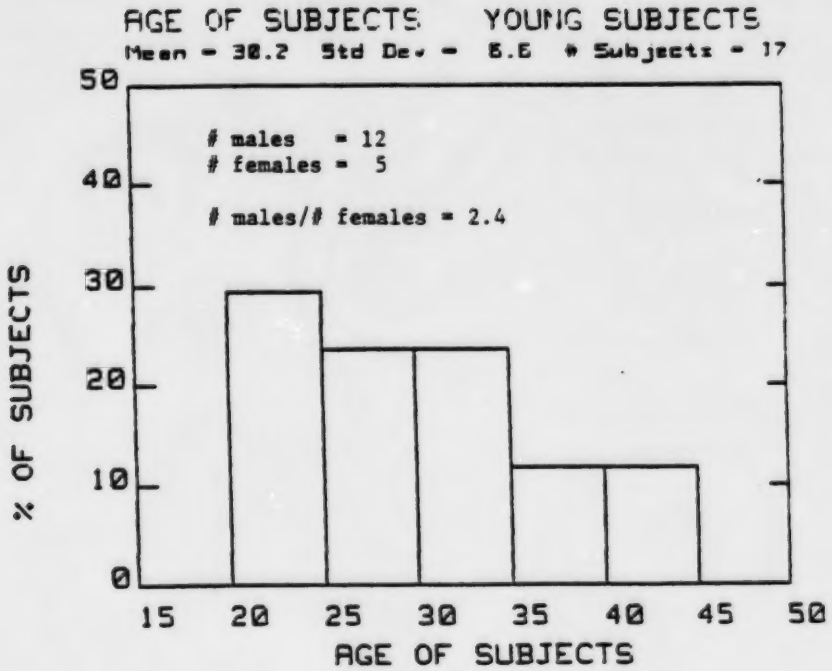


Figure 4.2 (a) Age Distribution of the Young Subjects

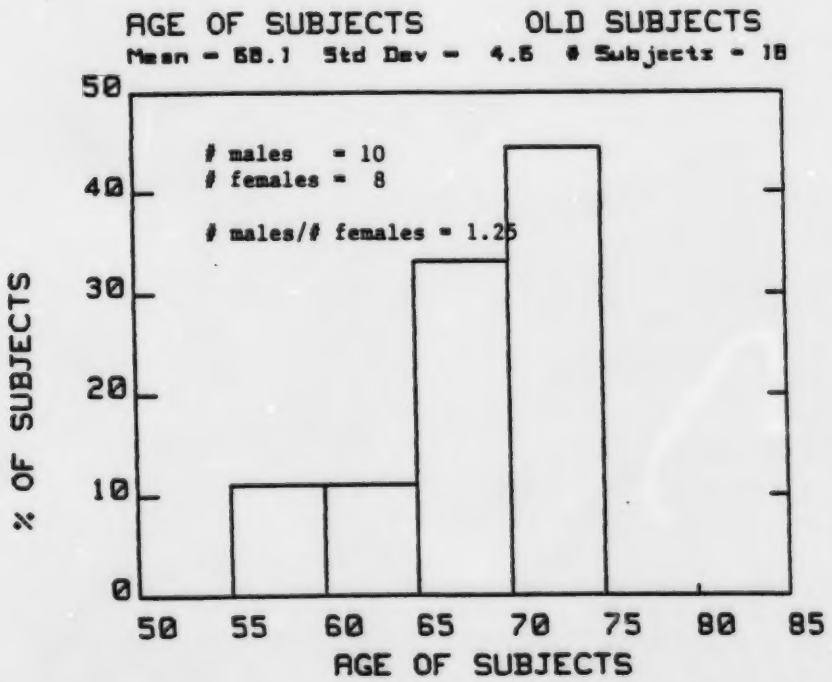


Figure 4.2 (b) Age Distribution of the Elderly Subjects

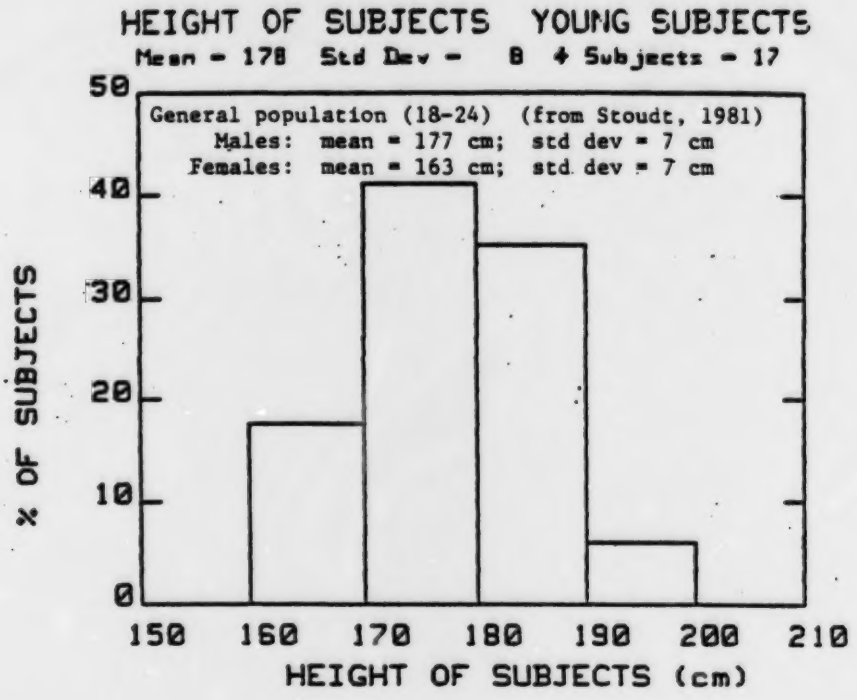

HEIGHT OF SUBJECTS YOUNG SUBJECTS
Mean = 178 Std Dev = 8 # Subjects = 17
General population (18-24) (from Stoudt, 1981)
Males: mean = 177 cm; std dev = 7 cm
Females: mean = 163 cm; std dev = 7 cm
% OF SUBJECTS
0
10
20
30
40
50
150 160 170 180 190 200 210
HEIGHT OF SUBJECTS (cm)

Figure 4.3 (a) Height Distribution of the Young Subjects

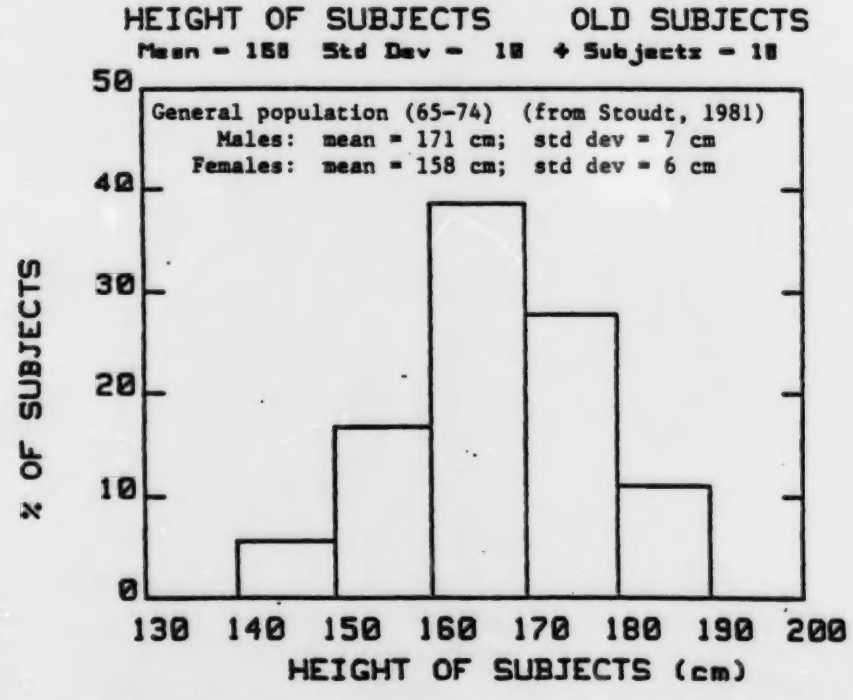

HEIGHT OF SUBJECTS OLD SUBJECTS
Mean = 158 Std Dev = 18 # Subjects = 18
General population (65-74) (from Stoudt, 1981)
Males: mean = 171 cm; std dev = 7 cm
Females: mean = 158 cm; std dev = 6 cm
% OF SUBJECTS
0
10
20
30
40
50
130 140 150 160 170 180 190 200
HEIGHT OF SUBJECTS (cm)

Figure 4.3 (b) Height Distribution of the Elderly Subjects

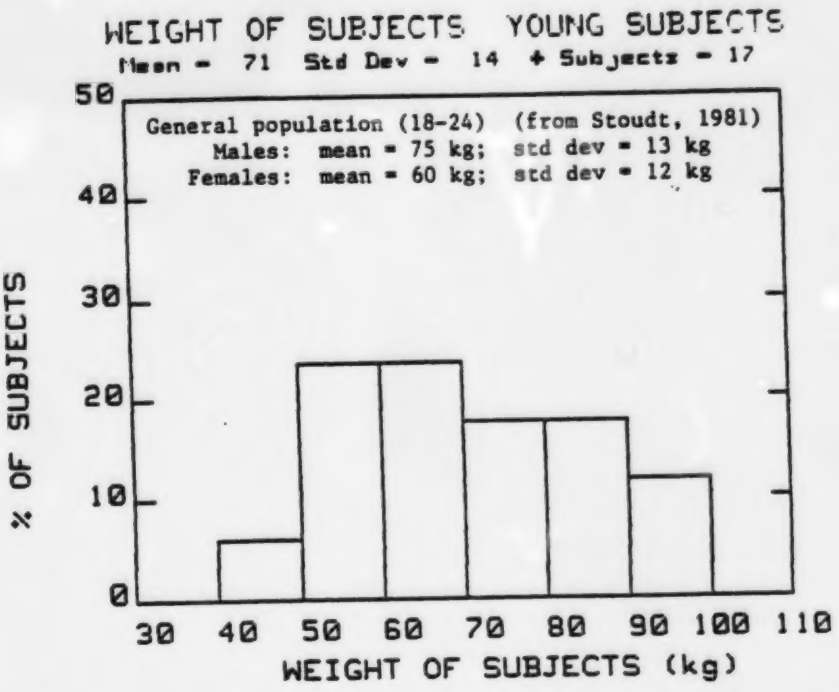

WEIGHT OF SUBJECTS YOUNG SUBJECTS
Mean = 71 Std Dev = 14 # Subjects = 17
General population (18-24) (from Stoudt, 1981)
Males: mean = 75 kg; std dev = 13 kg
Females: mean = 60 kg; std dev = 12 kg
% OF SUBJECTS
0
10
20
30
40
50
30 40 50 60 70 80 90 100 110
WEIGHT OF SUBJECTS (kg)

Figure 4.4 (a) Weight Distribution of the Young Subjects

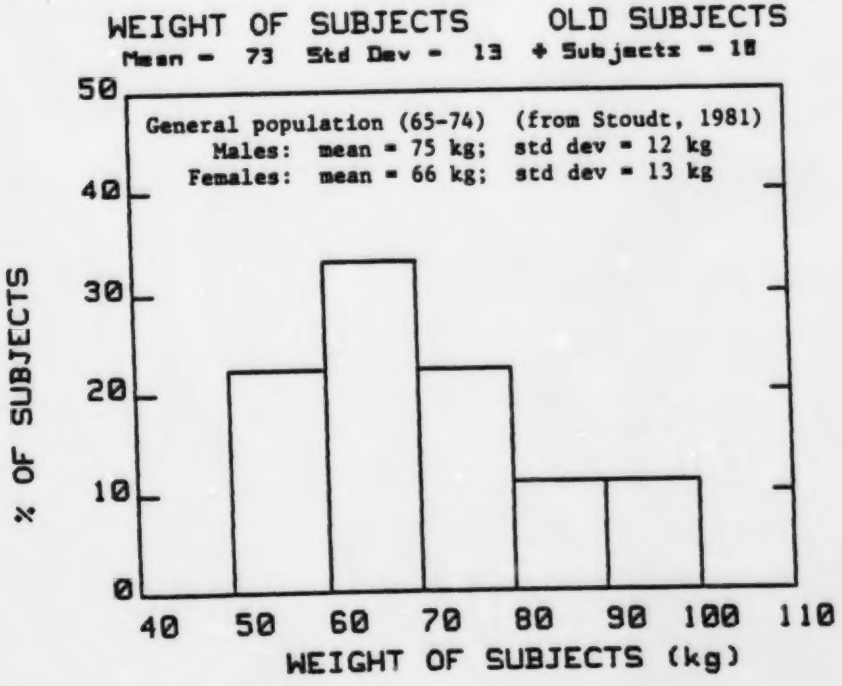

WEIGHT OF SUBJECTS OLD SUBJECTS
Mean = 73 Std Dev = 13 # Subjects = 18
General population (65-74) (from Stoudt, 1981)
Males: mean = 75 kg; std dev = 12 kg
Females: mean = 66 kg; std dev = 13 kg
% OF SUBJECTS
0
10
20
30
40
50
40 50 60 70 80 90 100 110
WEIGHT OF SUBJECTS (kg)

Figure 4.4 (b) Weight Distribution of the Elderly Subjects

Experimental Apparatus

Photographs of the experimental apparatus are shown in Figures 4.5 and 4.6. As mentioned earlier, the subject stands on a mock stairway, so that he/she is able to judge his/her preferred lateral position and grip location. The subject is restrained posteriorly by a rigid, foam-cushioned back-board, and anteriorly by a canvas harness. The restraining backboard is mounted at the top of the mock stairway. A scale built into the tread of the stairway allows foot position to be recorded (see Figure 4.7).

The lateral displacement of the subject is adjusted by moving the mock stairway. Tracks ensure that anterior-posterior alignment of the stairway and handrail are maintained as lateral position is adjusted. The mock stairway is clamped to the tracks to prevent motion during the experimental tests. A built-in scale allows measurement of lateral displacement.

As shown in Figure 4.8, the handrail support is mounted on a cantilevered column, which is bolted to a piezoelectric force plate [Kistler, Model 9281]. The support and column were designed to provide an acceptable degree of rigidity to the handrail. With a 500 Newton (100 lb) load applied at the extreme end of the handrail, the actual deflection of the handrail is approximately 25 mm (1/4 inch). This degree of compliance did not appear to affect the performance of the subjects.

The height and slope of the handrail are both adjustable. Built-in scales allow for convenient measurement of these parameters. The handrail support is mounted on a rack and pinion, to facilitate changes in height, as shown in Figure 4.9. In addition to a friction clamp, a locating pin and set screw are used to fix the support to the column at each height, thereby eliminating any motion of the handrail and ensuring that the height can be set easily and repeatably at the correct levels. The handrail itself is removable so that different shapes, sizes and materials can be tested if desired, at some future date. Handrail height, slope and diameter can all be adjusted independently of each other.

The force plate measures horizontal forces, vertical force, and all three bending moments. The output of the force plate is fed to charge amplifiers [Kiag Swiss, Model 5001], and is then sampled using an analog-to-digital converter unit [Neff, Model 920] and a minicomputer [Hewlett Packard, Model 9845]. Sampling is performed at 20 Hz. Since the frequency content of the signal lies well below the aliasing frequency (10 Hz), this sampling rate is more than adequate. (The absence of aliasing was confirmed through experimentation with pre-aliasing filters.) The sampled data is stored on magnetic disc, for later data analysis.

The lack of dc response in the force plate/charge amplifier negates the loading due to the weight of the apparatus without affecting measurement of the forces and moments generated by the subject. During the tests, the charge amplifiers are periodically re-set, to prevent drift. In addition, the output is zeroed prior to each measurement.

Using the measured foot location on the stair tread and the known force plate-tread distance, the stabilizing moment (about the axis of rotation during "fixed-axis" falls, i.e. the metatarsophalangeal joint of the foot) is calculated from the measured forces and moments. Thus, stabilizing moment is recorded directly, without the need to measure moment arms. Note that the location of the grip on the handrail can also be calculated from the measured data, if it becomes necessary to study this parameter (however, the accuracy of this calculation deteriorates as the line of action of the handrail force and the slope of the handrail become parallel).

In the present experiment, anterior-posterior body motion was of prime concern; therefore, only three parameters were measured: (1) anterior-posterior horizontal force, (2) vertical force and (3) anterior-posterior moment (about the metatarsophalangeal joint). The computer software plots out the time records of these three parameters following each experimental trial (i.e. each instruction to the subject), so that the trial can be repeated if necessary.

The experimental apparatus was calibrated by hanging weights at known moment arms. Horizontal and upward vertical forces were achieved by means of a pulley arrangement. The results of the calibration are shown in Appendix A. Sources of error include non-linearity and hysteresis of the force plate, and cross-talk between the different force plate channels. Note that at least some of the error is associated with measurement of the moment arms, and is not attributable to the apparatus itself. Nonetheless, the discrepancies are generally less than 10%, and the linearity between applied and measured loads is generally quite good. Note that in comparative measurements (as were made in this study), the relative accuracy and linearity of the measurements are of greater concern than the absolute accuracy.

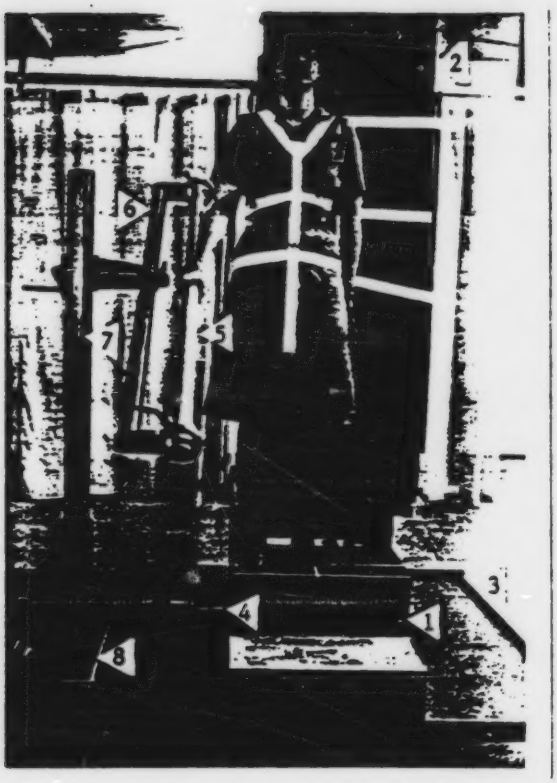


Figure 4.5 Frontal View of the Experimental Apparatus

Legend:
1. mock stairway
2. restraining back-board
3. tracks (for adjusting lateral displacement)
4. scale for measuring lateral displacement
5. handrail
6. support for handrail
7. scale for measuring height of handrail
8. force plate

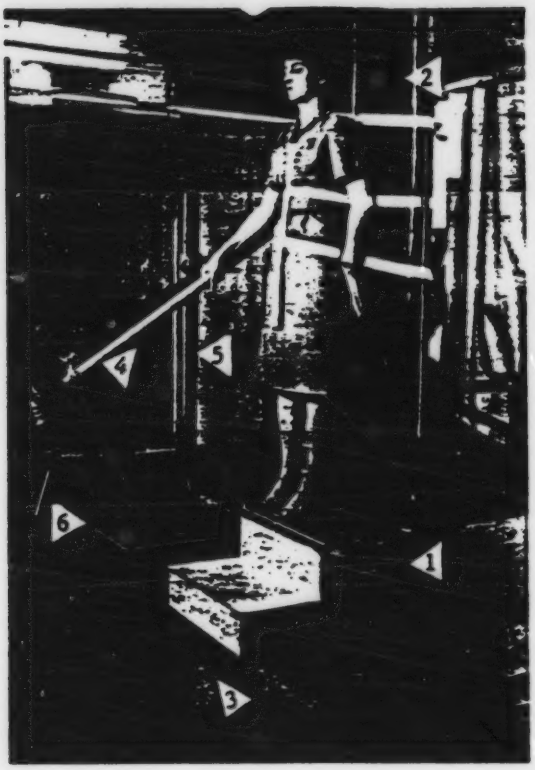


Figure 4.6 Side View of the Experimental Apparatus

Legend:
1. mock stairway
2. restraining back-board
3. tracks (for adjusting lateral displacement)
4. handrail
5. support for handrail
6. force plate

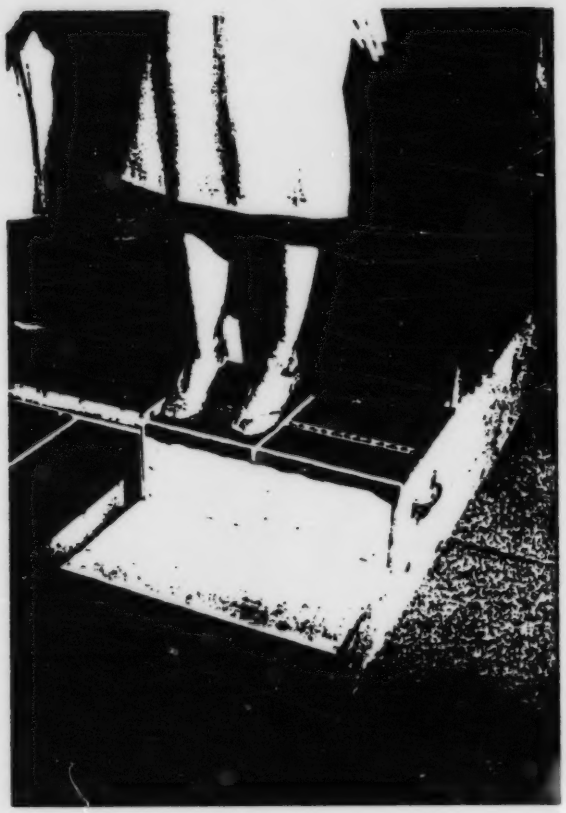


Figure 4.7 The Stair Tread

(Note: scale for measuring foot position)

Figure 4.8 The Handrail Support

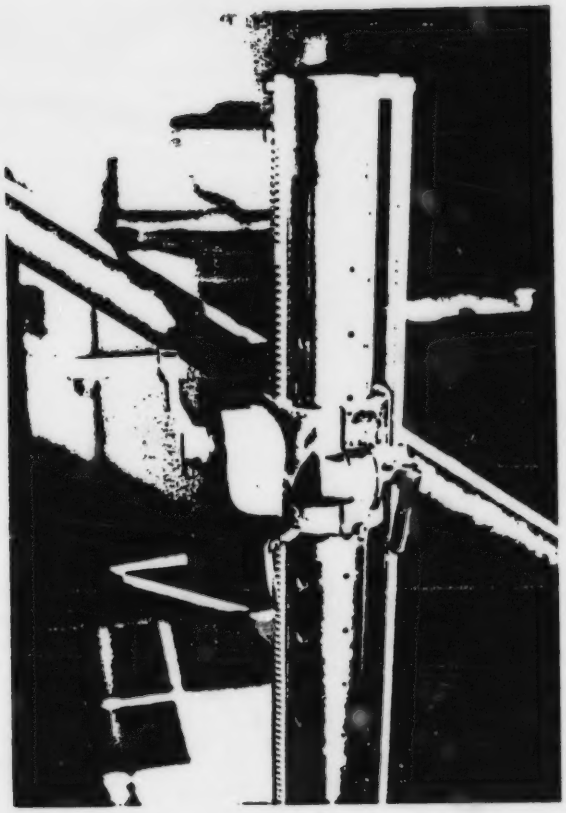


Figure 4.9 Height Adjustment of the Handrail

Details of the Experimental Protocol

The details of the protocol are as follows:

(1) The subject's age and sex are recorded, and appropriate anthropometric measurements are made (i.e. weight, height, shoulder height, arm span, and arm and hand dimensions). Pertinent medical data is also recorded (i.e. history of falling, heart disease, musculoskeletal disorders or injuries, medications, athletic activity, and somatotype).

(2) The handrail height is set to 36" (mid-range), the mock stairway is moved to the position of minimum displacement, and the subject is asked to choose his/her preferred lateral displacement. The subject's foot position is recorded, and the required position of the stairway is calculated. The lateral displacement between the body and handrail centre-lines is calculated and recorded.

(3) The stairway is moved to the appropriate lateral location, and the subject is asked to stand at the right-hand edge of the stairway. The restraining harness is adjusted to fit the subject and is then used to strap the subject to the back-board.

(4) The subject is asked to decide where to place his/her feet, and the location of the feet is marked on the tread using masking tape. The subject is asked to stand on his/her toes, and the approximate location of the metatarsophalangeal joint is measured.

(5) Starting at a randomly-selected handrail height, the subject is instructed to imagine that he/she is descending the stairway and to select his/her grip location on the handrail. The position is marked with masking tape and is measured and recorded.

(6) The force plate output is zeroed with the subject's hand resting on the handrail. The subject is then instructed to begin to push on the handrail (forward along the handrail), to gradually increase the force to his/her maximum capability and then to relax. The test begins at the sound of a beep; a second beep sounds after four seconds. The subject is instructed to relax at the sound of the second beep, if he/she has not already relaxed at that point. The forces and moments generated at the force plate are automatically sampled and stored by the minicomputer during each test. One minute of rest is allowed, during which the subject is allowed to move his/her feet and arms. During this time, the experimenter examines the results on the computer terminal. If the results indicate that the subject misunderstood the instruction or if some other problem occurred, then the test is repeated.

(7) Step (6) is repeated with instructions to "pull backwards along the handrail", "push downwards towards the ground", and "pull upwards away from the ground". Prior to each test, the experimenter ensures that the subject's hand and foot locations have not changed.

(8) Steps (5), (6) and (7) are repeated at each handrail height, selected in random order. After each height is tested, the subject is asked to indicate his preferred height (to that point). The charge amplifier is also re-set after completing each height. After four heights have been tested, the subject is given the option of taking a rest.

(9) Steps (5), (6) and (7) are repeated at the original handrail height, using the same grip location as was used during the original trial.

Methods of Analysis

(1) Introduction

The goals of the data analysis are: (1) determine whether the ability to generate stabilizing forces and moments depends significantly on handrail height; (2) if the effect of handrail height is significant, determine the nature of the relationship between handrail height and force/moment generation; (3) determine whether there are significant age-related differences in either the ability to generate stabilizing forces and moments or in the dependence of this ability on handrail height; and (4) estimate a range of acceptable handrail heights for use in building design.

(2) Parameters

The stabilizing forces and moments (acting on the body) measured under the various instructions can be related to the different types of falling motion, as follows:

1. forward rotation - maximum backward moment (P1)
2. forward translation - maximum backward force (P2)
3. downward translation - maximum upward force (P3)
4. backward rotation - maximum forward moment (P4)
5. backward translation - maximum forward force (P5)

These five forces and moments are used to characterize the performance of each subject at each handrail height. Note that the variable names listed in brackets (P1, P2, etc.) are used throughout the remainder of the report in referring to the five parameters. The above falling motions and corresponding forces and moments are listed in order of estimated importance during stairway descent. Thus, for example, forward rotation is judged to have the most serious implications for injury to the faller. Accordingly, backward moment is judged to be the most important parameter in evaluating the effectiveness of a given handrail.

Data from a typical trial is shown in Figure 4.10. As is evident in this data, significant amounts of intention tremor (2-4 Hz) can appear in the force and moment data. To eliminate this, as well as other sources of noise, the raw data is passed through a twelfth-order, zero phase-lag digital low-pass filter, with cut-off at 1 Hz.

From the filtered data records, maximal values of the five stabilizing forces and moments are found for each experimental trial (i.e. each instruction). From the four trials conducted at each handrail height, overall maximal values are selected. Subsequent analysis uses these maximal values to characterize each subject's performance at each handrail height.

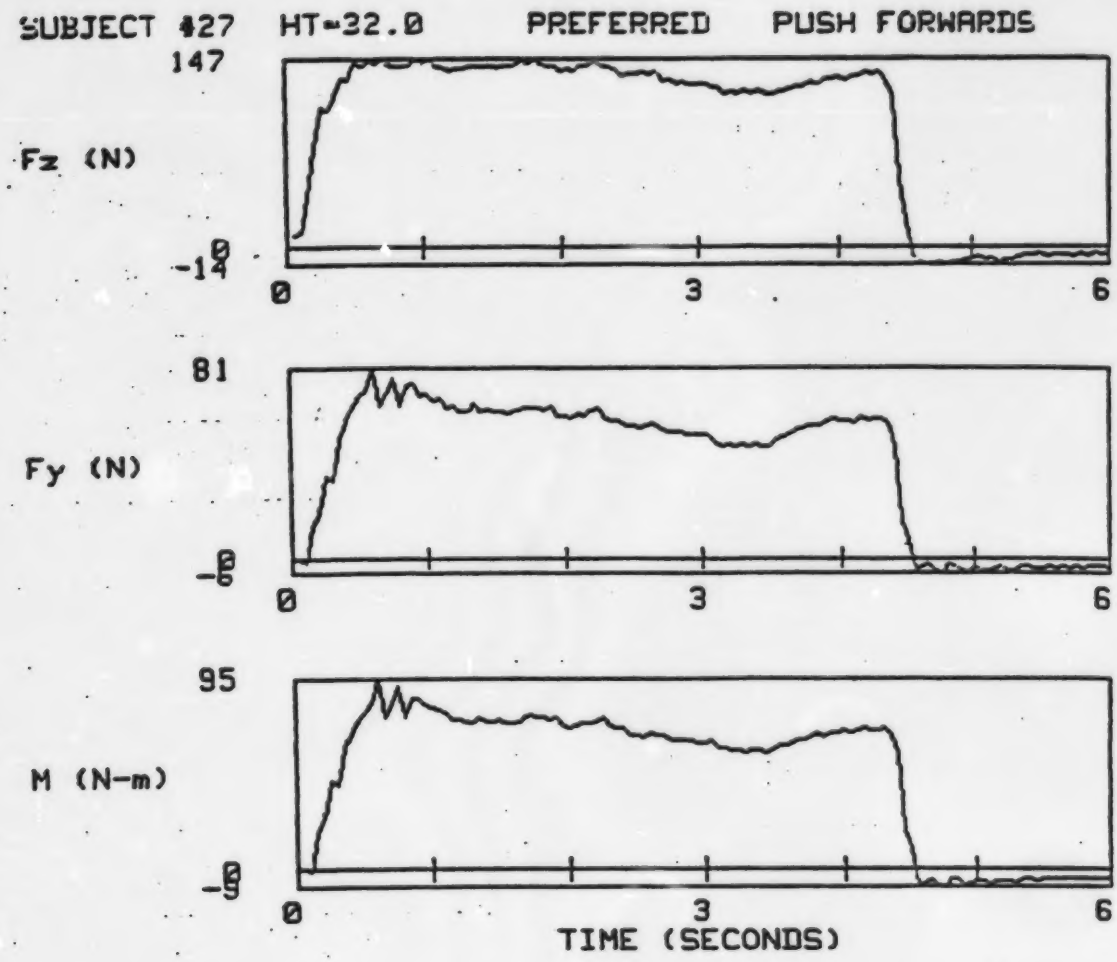


Figure 4.10 Data from a Typical Experimental Trial

Note: Fz = vertical force
Fy = horizontal force
M = moment (about MP joint)

(3) Non-Dimensionalization of the Parameters

Use of absolute force and moment values is meaningless in comparing different individuals, since the stabilizing effect of a given force or moment is dependent on both body weight and size. In order to compare different subjects, the force and moment parameters must be non-dimensionalized in an appropriate way.

One approach is as follows: for each subject, divide all values of a given parameter by the maximum value of that parameter obtained by that particular subject. This would result in values ranging from zero to one, with each value representing the relative effectiveness of a given handrail height for a particular subject in generating a particular type of force or moment. The major problem with this approach is that the non-dimensionalized data no longer has a normal distribution, and hence standard statistical tests cannot be applied.

The approach used here was to divide each subject's forces by his/her body weight, and to divide moments by the "body weight moment" (a factor proportional to the product of body weight and height). This approach can be justified in at least two different ways. During a fall, body weight acts as the destabilizing force, whereas the destabilizing moment is related to the product of body weight and height. Thus, each non-dimensionalized value indicates the stabilizing effect of the handrail relative to the corresponding destabilizing factor. From a purely statistical point of view, the variance of a particular force or moment (at the different handrail heights) must be approximately equal for all of the subjects, in order to perform a statistical comparison. Non-dimensionalization by body parameters was found to have the effect of equalizing "within-subject" variance.

Note that the actual destabilizing moment is proportional to the sine of the angle of the body (relative to vertical). The maximum possible destabilizing moment was used for non-dimensionalization. This value is equal to the product of body weight times the distance between the centre-of-gravity and the metatarsophalangeal joint. For the analysis, this distance was estimated to be 0.53 times body height [Roebuck, 1975] (sex differences in the location of the centre-of-gravity were neglected).

In all discussion that follows, non-dimensionalization with respect to body parameters is assumed, unless specified otherwise. Thus, the parameters P1 to P5 are assumed to refer to non-dimensionalized values.

(4) Analysis of Variance - Split Plot Design

Analysis of variance (with split plot design) was used to evaluate the effects of age and handrail height on the generation of stabilizing forces and moments. In using this statistical model, the two age groups are considered to be separate plots, the subjects are considered to be separate blocks within each plot, and the handrail heights are considered to be the treatments.

(5) Analysis of Covariance

Analysis of covariance was used to investigate the nature of the relationship between handrail height and ability to generate stabilizing forces and moments. This statistical model assumes a relationship of the form:

$$Yij = u + Ai + Bi\ Xj + Eij$$

where:

- Yij = parameter value for ith subject at the jth handrail height
- u = overall mean effect
- Ai = y-intercept of ith subject (minus u)
- Bi = slope of ith subject
- Xj = jth handrail height
- Eij = error term (due primarily to motivational variation)

Qualitatively, the linear model can be justified through examination of the data (see Figures 4.11 to 4.15), which shows a pronounced linear trend. Quantitatively, attempts to perform higher-order polynomial regression on the data showed negligible improvement over a linear fit.

Use of this model allows the following hypotheses to be tested, using F-statistics: (1) the slopes of all subjects are equal, (2) the slopes of all subjects are equal to zero, and (3) the y-intercepts of all subjects are equal. Note that the slope (Bi) represents the dependence of non-dimensionalized force or moment generation on handrail height, whereas the y-intercept (Ai) is indicative of the "relative magnitude" of non-dimensionalized force or moment generated by each subject.

(6) Estimation of an Optimal Design Range

In theory, for each parameter, a regression versus handrail height could be performed. Then, by constructing a 90% confidence interval about the regression line and using the lower confidence limit, handrail heights could be selected which would allow 95% of the user population to generate some given fraction of their body weight or body weight moment. In practice, the data did not allow this approach to be used. As will be seen in the next section, the y-intercepts of the different subjects are significantly different. Thus, even though the slopes are the same, the confidence interval on the regression line is too wide to allow a meaningful handrail height design range to be specified.

Another approach that was explored involved plotting each non-dimensionalized parameter versus handrail height for the two extreme individuals in each age group (i.e. those individuals who recorded maximal force or moment at handrail heights lower and higher

than any other subjects). In theory, an acceptable range for handrail height could be estimated from the overlap between the two plots, as was illustrated earlier in Figure 2.3. In practice, the separation between individuals was not sufficiently clear-cut and no such extreme individuals could be identified from the data.

The approach that was eventually used involved constructing histograms for each of the five parameters (treating the two age groups separately). The histograms represent the percentage of the experimental subjects (in each age group) who were able to generate at least some given fraction of their body weight (or body weight moment) at each handrail height. For each of the five parameters, the critical fraction was chosen so that at least 75% of the subjects were able to generate this fraction of their body weight (or body weight moment) for at least four of the six handrail heights. Thus, analysis of each parameter yields at least four handrail heights that are considered to be acceptable for at least 75% of the subjects. By comparing the results obtained for the five different parameters, a suitable design range is obtained.

(7) Other Analyses

In addition to the analyses described above, possible correlations between performance and body height, weight, and armspan were explored. Such correlations, if they existed, would be useful in predicting the optimal handrail height for any given user population, by using available anthropometric data. This analysis involved performing a regression of each non-dimensionalized stabilizing force or moment versus the particular body parameter of interest, for each handrail height. Unfortunately, the relatively small number of experimental subjects resulted in very low confidence in the regression, and precluded any meaningful conclusions. This approach might be successful if more data were available.

Results

The data is plotted in Figures 4.11 to 4.15. Each plot represents the variation of a given parameter with handrail height. The result for each subject is plotted separately, to allow trends to be identified. To facilitate comparison, results for the young and the elderly subjects have been plotted on separate graphs. The data in these plots has been non-dimensionalized with respect to body parameters, as described above.

(1) Effect of Handrail Height

(i) Difference between Means

The results of the analysis of variance (split plot design) are summarized in Table 4.1. The analysis of the five parameters was performed separately. For each of the five parameters, the mean parameter value was calculated at each of the six handrail heights, using all of the subjects. Associated 95% confidence limits were also calculated for each mean.

The null hypothesis states that the means at the different handrail heights are equal. The level of significance indicated in the table refers to the test of the null hypothesis. Small levels of significance imply that the null hypothesis is incorrect, namely, that there is a significant difference between the means.

It is evident from the results shown in the table that the differences between the means are highly significant. This is true for all five parameters. Thus, we conclude that handrail height has a highly significant effect on the ability to generate stabilizing forces and moments.

Note that for parameters P1, P2, P4 and P5, the mean values increase with increasing handrail height. For parameter P3 (vertical force), the mean values decrease with increasing handrail height. These trends are also evident in the "raw" data (Figures 4.11 to 4.15).

(ii) Linear Dependence

The nature of the relationship between the parameter values and handrail height was evaluated using analysis of covariance. From the plots in Figures 4.11 to 4.15, we identified a particular subject who consistently generated much higher non-dimensionalized forces and moments than the other subjects. This subject apparently was able to use his body weight in augmenting his handrail force, in spite of the restraining harness. This subject was excluded from the analysis of covariance. The results of the analysis are summarized in Table 4.2.

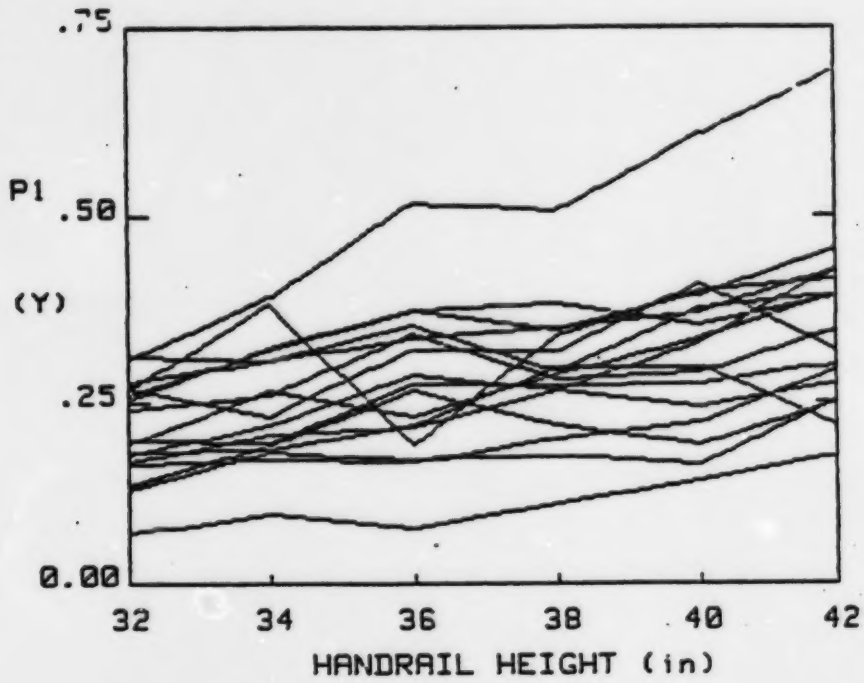


Figure 4.11 (a) Parameter P1 versus Handrail Height - Young Subjects
(P1 = maximum backward moment, non-dimensionalized)

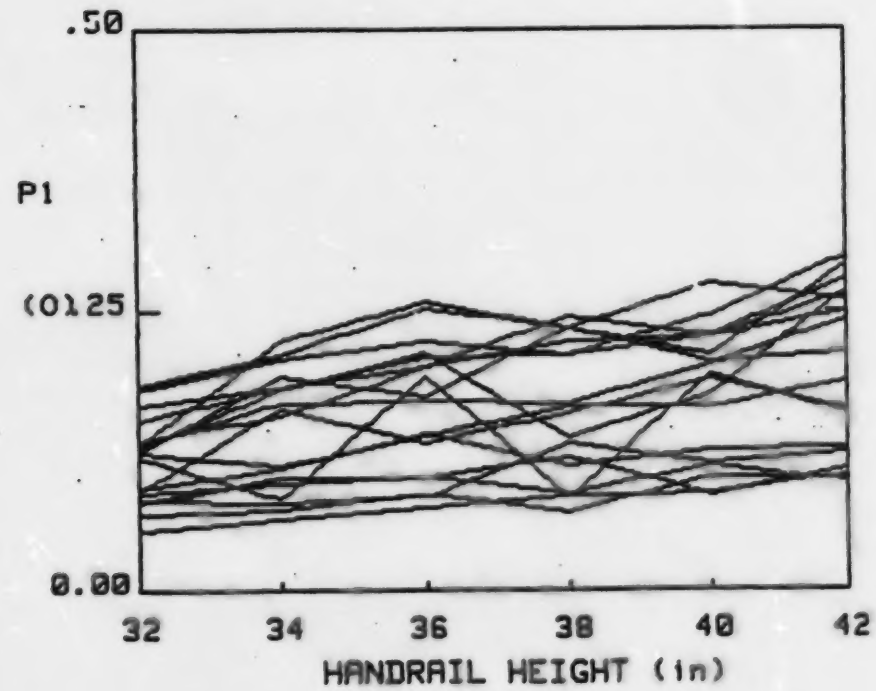


Figure 4.11 (b) · Parameter P1 versus Handrail Height - Elderly Subjects
(P1 = maximum backward moment, non-dimensionalized)

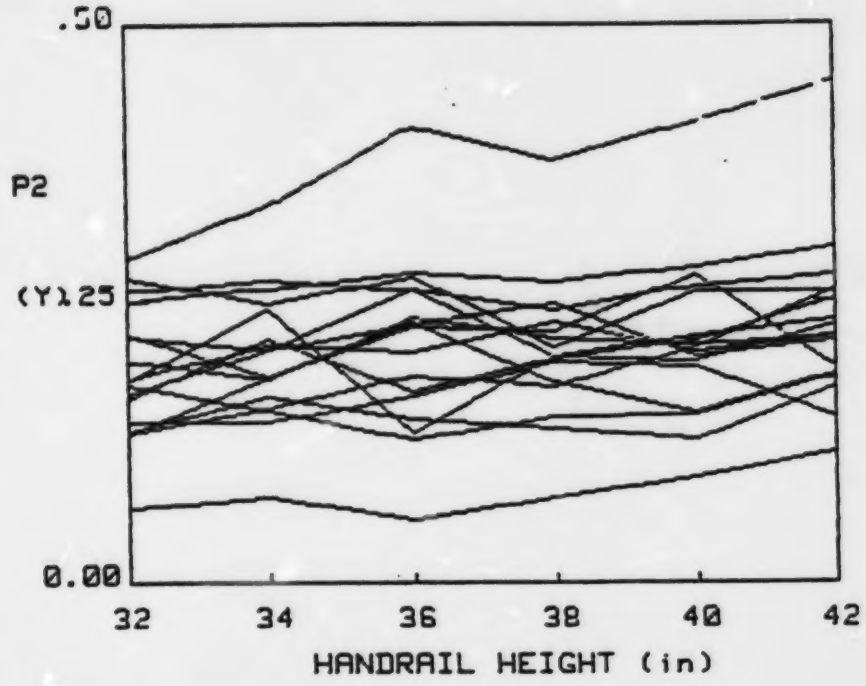


Figure 4.12 (a) Parameter P2 versus Handrail Height - Young Subjects
(P2 = maximum backward force, non-dimensionalized)

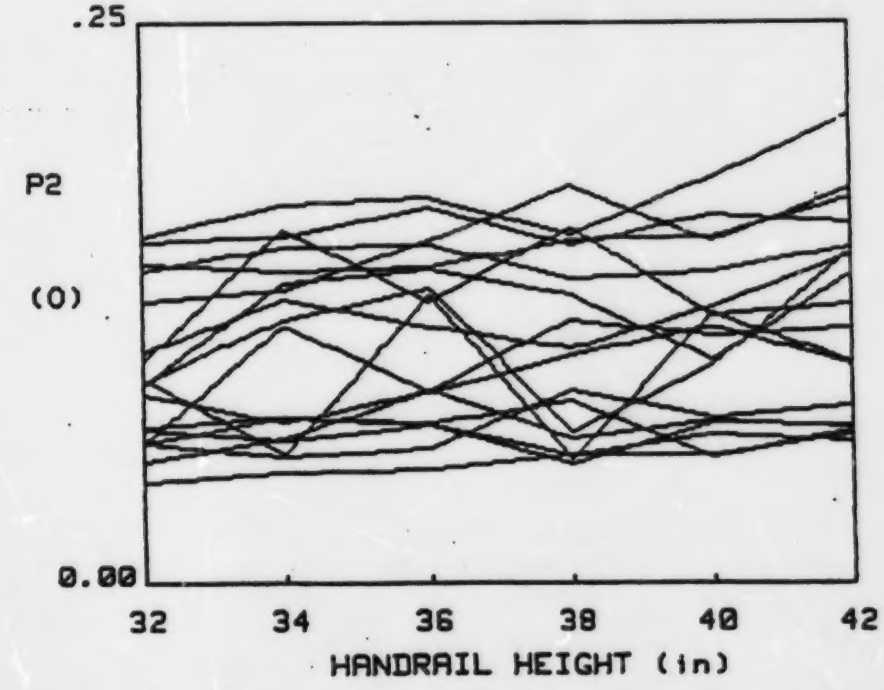


Figure 4.12 (b) Parameter P2 versus Handrail Height - Elderly Subjects
(P2 = maximum backward force, non-dimensionalized)

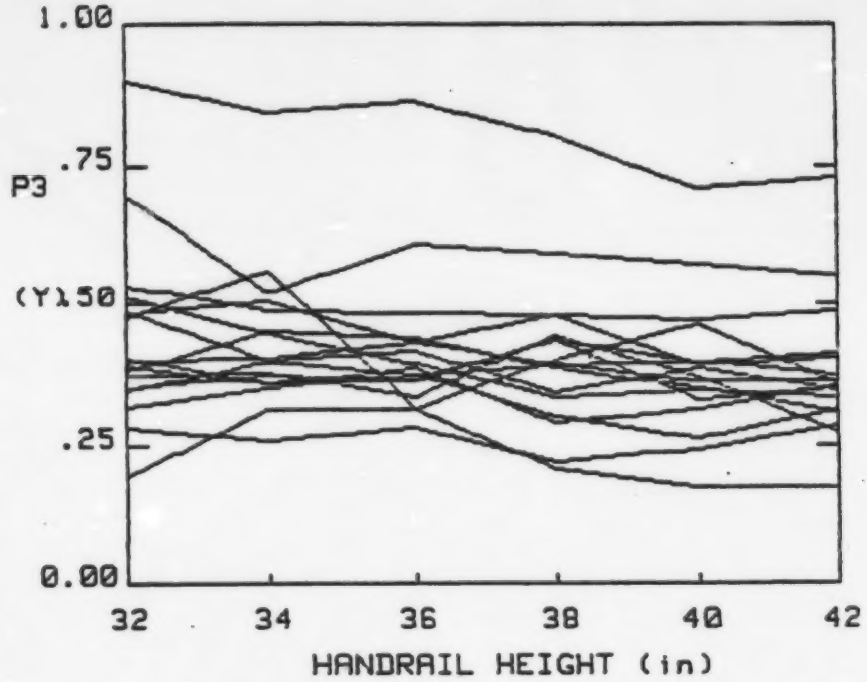


Figure 4.13 (a) Parameter P3 versus Handrail Height - Young Subjects
(P3 = maximum upward force, non-dimensionalized)

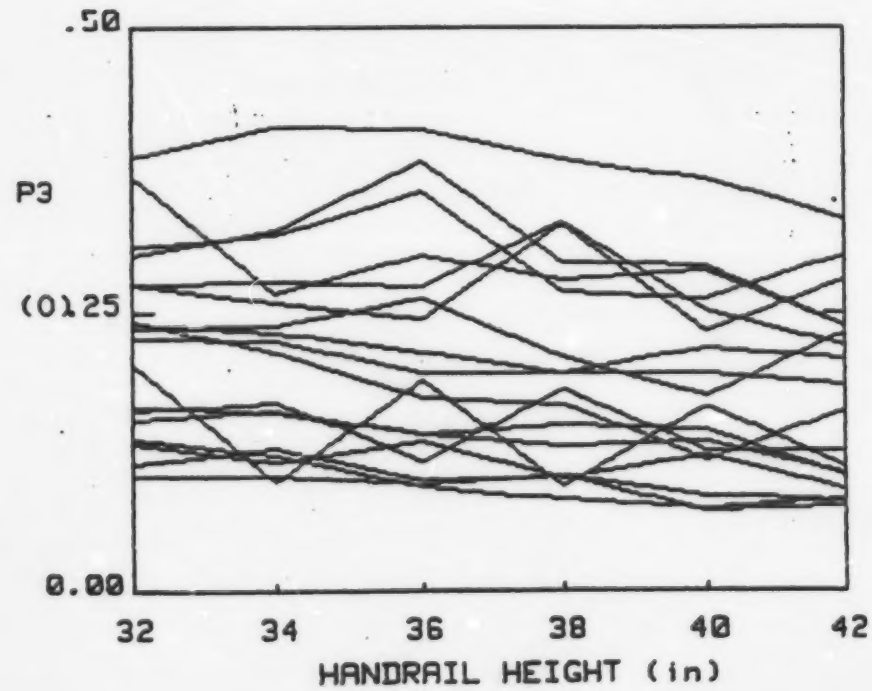


Figure 4.13 (b) Parameter P3 versus Handrail Height - Elderly Subjects
(P3 = maximum upward force, non-dimensionalized)

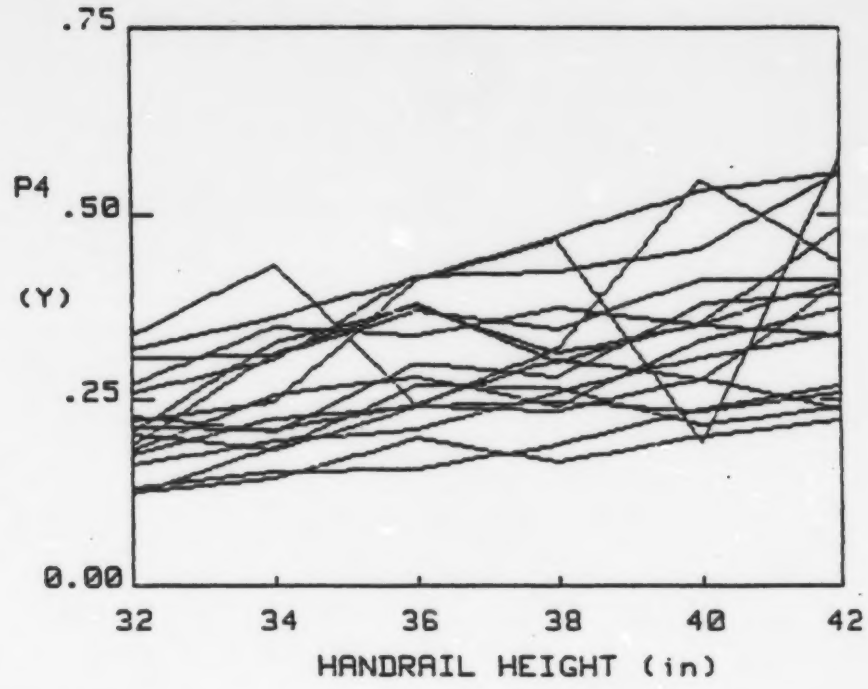


Figure 4.14 (a) Parameter P4 versus Handrail Height - Young Subjects
(P4 = maximum forward moment, non-dimensionalized)

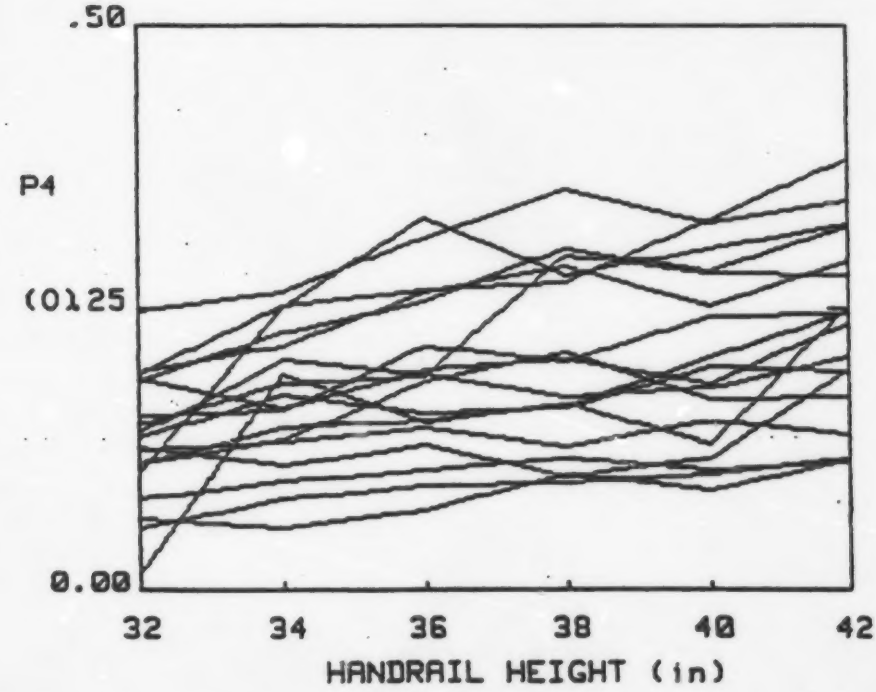


Figure 4.14 (b) Parameter P4 versus Handrail Height - Elderly Subjects
(P4 = maximum forward moment, non-dimensionalized)

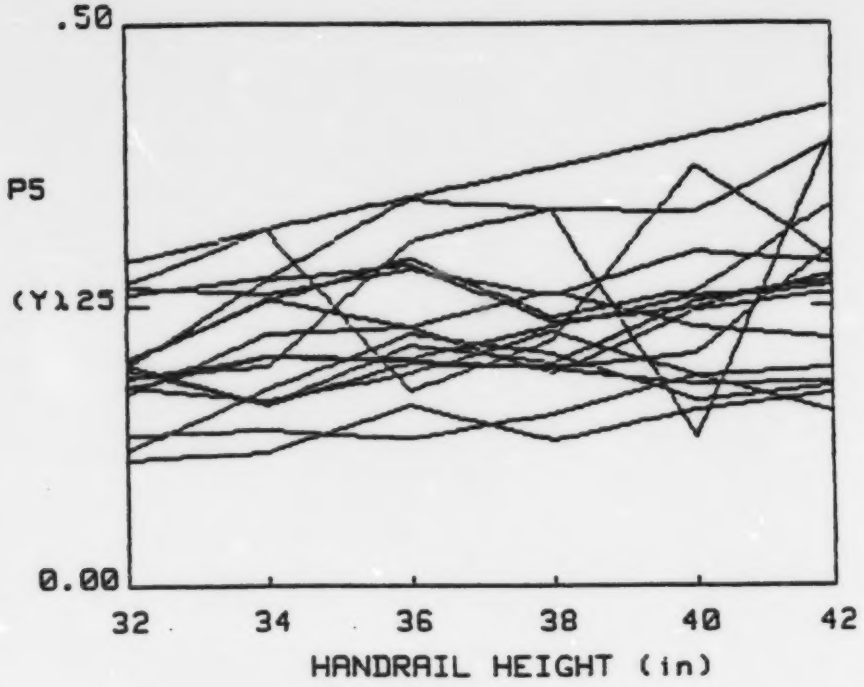


Figure 4.15 (a) Parameter P5 versus Handrail Height - Young Subjects
(P5 = maximum forward force, non-dimensionalized)

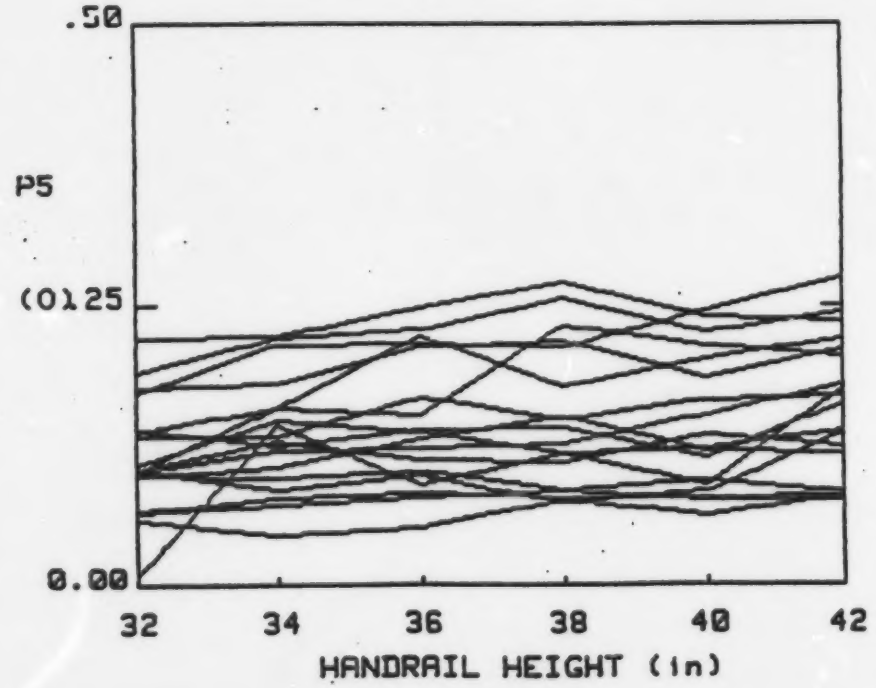


Figure 4.15 (b) Parameter P5 versus Handrail Height - Elderly Subjects
(P5 = maximum forward force, non-dimensionalized)

TABLE 4.1 - Effect of Handrail Height

	Handrail Height	Mean	95% Confidence Interval	
Parameter P1	32"	0.163	0.152	0.174
	34"	0.193	0.181	0.204
(Backward	36"	0.217	0.206	0.228
Moment)	38"	0.221	0.210	0.232
	40"	0.243	0.232	0.254
	42"	0.271	0.260	0.282
	Level of significance: 0.00000			
	Conclusion: significant difference between means			
Parameter P2	32"	0.137	0.129	0.144
	34"	0.151	0.144	0.159
(Backward	36"	0.159	0.151	0.166
Force)	38"	0.154	0.147	0.161
	40"	0.161	0.153	0.169
	42"	0.172	0.165	0.179
	Level of significance: 0.00000			
	Conclusion: significant difference between means			
Parameter P3	32"	0.325	0.313	0.338
	34"	0.316	0.304	0.328
(Upward	36"	0.310	0.297	0.322
Force)	38"	0.297	0.285	0.309
	40"	0.278	0.266	0.290
	42"	0.270	0.258	0.282
	Level of significance: 0.00000			
	Conclusion: significant difference between means			
Parameter P4	32"	0.172	0.158	0.186
	34"	0.211	0.197	0.226
(Forward	36"	0.240	0.225	0.254
Moment)	38"	0.252	0.238	0.267
	40"	0.262	0.248	0.277
	42"	0.302	0.288	0.317
	Level of significance: 0.00000			
	Conclusion: significant difference between means			
Parameter P5	32"	0.154	0.144	0.164
	34"	0.174	0.164	0.184
(Forward	36"	0.187	0.177	0.197
Force)	38"	0.189	0.179	0.199
	40"	0.190	0.180	0.200
	42"	0.214	0.203	0.224
	Level of significance: 0.00000			
	Conclusion: significant difference between means			

TABLE 4.2 - Linear Dependence on Handrail Height

Parameter P1 - Backward Moment: Pooled slope = 0.00922 R = 0.9423

(1) Ho: All subjects have the same slope.
Level of significance = 0.02873
Conclusion: Reject Ho at p ≥ .029
(2) Ho: Common slope is zero.
Level of significance = 0.00000
Conclusion: Reject Ho.
(3) Ho: All subjects have the same y-intercept.
Level of significance = 0.00000
Conclusion: Reject Ho.

Parameter P2 - Backward Force: Pooled slope = 0.00236 R = 0.9567

(1) Ho: All subjects have the same slope.
Level of significance = 0.56831
Conclusion: Accept Ho.
(2) Ho: Common slope is zero.
Level of significance = 0.00000
Conclusion: Reject Ho.
(3) Ho: All subjects have the same y-intercept.
Level of significance = 0.00000
Conclusion: Reject Ho.

Parameter P3 - Upward Force: Pooled slope =-0.00552 R =-0.9664

(1) Ho: All subjects have the same slope.
Level of significance = 0.36367
Conclusion: Accept Ho.
(2) Ho: Common slope is zero.
Level of significance = 0.00000
Conclusion: Reject Ho.
(3) Ho: All subjects have the same y-intercept.
Level of significance = 0.00000
Conclusion: Reject Ho.

Parameter P4 - Forward Moment: Pooled slope = 0.01139 R = 0.9042

(1) Ho: All subjects have the same slope.
Level of significance = 0.04824
Conclusion: Reject Ho at p ≥ .049.
(2) Ho: Common slope is zero.
Level of significance = 0.00000
Conclusion: Reject Ho.
(3) Ho: All subjects have the same y-intercept.
Level of significance = 0.00000
Conclusion: Reject Ho.

Parameter P5 - Forward Force: Pooled slope = 0.00486 R = 0.9245

(1) Ho: All subjects have the same slope.
Level of significance = 0.25392
Conclusion: Accept Ho.
(2) Ho: Common slope is zero.
Level of significance = 0.00000
Conclusion: Reject Ho.
(3) Ho: All subjects have the same y-intercept.
Level of significance = 0.00000
Conclusion: Reject Ho.

For each parameter, the correlation coefficient was calculated. As shown in the table, all five parameters had correlation coefficients with absolute values greater than 0.9. This implies that the linear model fits the data well; therefore, we conclude that there is a strong linear relationship between the parameter values and handrail height. The positive correlation coefficients obtained for parameters P1, P2, P4 and P5 indicate that these parameters increase with increasing handrail height. The negative value obtained for P3 indicates that this parameter (vertical force) decreases with increasing handrail height.

The pooled (common) slope was also calculated for each parameter. Recall that the slope is a measure of the linear dependence of force or moment generation on handrail height. Quantitatively, these values indicate the change in the non-dimensional parameter value per unit change in handrail height (in inches). Thus, for example, an increase of ten inches in handrail height would result in an increase of 0.0922 (10 x 0.00922) in parameter P1, on the average. This means that this increase in handrail height would enable the average subject to increase his/her backward moment by approximately 9% of his/her body weight moment. Note that the pooled slopes are all positive for parameters P1, P2, P4 and P5, but negative for parameter P3, thereby confirming the trends mentioned earlier.

It is evident from the plots (Figures 4.11 to 4.15) that while many subjects have approximately equal slopes, the y-intercepts are substantially different. The analysis of covariance tests both of these hypotheses. In addition, given equal slopes, it allows us to check that the common slope is not zero, which would imply no dependence of force/moment generation on handrail height.

The null hypothesis states that all subjects have the same slope. Small levels of signficance imply that the hypothesis is incorrect, or that the slopes are signficantly different. For parameters P2, P3, and P5, the levels of signficance are large; therefore, we accept the hypothesis that, for parameters P2, P3 and P5, all subjects have the same slope.

For parameters P1 and P4 (the two moments), the level of significance is borderline. Therefore, for these two parameters, further analysis was performed. One possibility is that the two age groups have different slopes but that all subjects within each age group have the same slope. This hypothesis was tested by repeating the analysis of covariance for each age group separately. The results showed a level of significance of 0.53157 for parameter P1, and 0.46885 for parameter P4, for the young subjects. For the elderly subjects, P1 had a level of significance of 0.07995 while P4 was 0.00892. These results imply that: (1) the young subjects have a common slope for parameter P1 and for parameter P4, and (2) the elderly subjects do not have a common slope for either parameter P1 or for parameter P4.

The hypothesis that the common slope is zero was found to be incorrect in all cases where there was a common slope. Thus, the earlier findings of a common slope are indicative of a significant dependence of the parameters on handrail height.

The hypothesis that all subjects have equal y-intercepts was found to be incorrect in all cases. This confirms the earlier qualitative finding.

(2) Age Differences

(i) Difference between Means

The results of the analysis of variance (split plot design) pertaining to age are listed in Table 4.3. For each parameter, mean values were calculated for each age group, based on results at all six handrail heights. Associated 95% confidence levels were also calculated for each mean.

The null hypothesis states that the means of the two age groups are equal. Small levels of significance indicate that the hypothesis is incorrect and that the means are in fact significantly different. As shown in the table, the results imply that there are significant age differences for all five parameters.

Examination of the data in the table shows that the mean values of the young subjects are higher than those of the elderly subjects.

(ii) Interaction between Age and Handrail Height

The interaction between age and handrail height is shown in Table 4.4. Significant interaction means that the nature of the dependence of force/moment generation on handrail height is different for the two age groups. For a linear dependence (as was found to exist here), this means that the two age groups should have different slopes.

The null hypothesis is that there is no signficant interaction; therefore, small levels of significance indicate that the hypothesis is wrong and that there is significant interaction. The results listed in the table indicate significant interaction for parameters P1 and P4; there is no significant interaction for parameters P2, P3 or P5.

The interaction results are confirmed by the analysis of covariance. The analysis of covariance showed that all subjects had common slopes for parameters P2, P3 and P5; therefore, there are no age difference in the slope for these parameters. However, for parameters P1 and P4 (the two moments), the young subjects were found to have a common slope whereas the elderly subjects did not.

TABLE 4.3 - Effect of Age

	Age Group	Mean	95% Confidence Interval	
Parameter P1				
(Backward Moment)	Young	0.276	0.229	0.322
	Elderly	0.160	0.114	0.207
	Level of significance: 0.00173 Conclusion: significant difference between means at $p \geq .002$			
Parameter P2				
(Backward Force)	Young	0.203	0.171	0.234
	Elderly	0.108	0.078	0.139
	Level of significance: 0.00027 Conclusion: significant difference between means			
Parameter P3				
(Upward Force)	Young	0.397	0.329	0.464
	Elderly	0.202	0.134	0.270
	Level of significance: 0.00049 Conclusion: significant difference between means			
Parameter P4				
(Forward Moment)	Young	0.292	0.249	0.334
	Elderly	0.188	0.146	0.231
	Level of significance: 0.00210 Conclusion: significant difference between means at $p \geq .003$			
Parameter P5				
(Forward Force)	Young	0.228	0.194	0.261
	Elderly	0.142	0.108	0.175
	Level of significance: 0.00131 Conclusion: significant difference between means at $p \geq .002$			

TABLE 4.4 - Interaction of Age and Handrail Height

Parameter P1 - Backward Moment

Level of significance: 0.00193
Conclusion: significant interaction at $p \geq .002$

Parameter P2 - Backward Force

Level of significance: 0.42150
Conclusion: no significant interaction

Parameter P3 - Upward Force

Level of significance: 0.81509
Conclusion: no significant interaction

Parameter P4 - Forward Moment

Level of significance: 0.03604
Conclusion: significant interaction at $p \geq .036$

Parameter P5 - Forward Force

Level of significance: 0.42984
Conclusion: no significant interaction

(3) Estimation of an Optimal Design Range for Handrail Height

The design histograms are plotted in Figures 4.16 to 4.20. These histograms show the percentage of subjects who were able to generate "acceptable" force or moment levels versus handrail height. Note that each parameter is analyzed separately, and that each age group is analyzed separately.

For each parameter (and for each age group), a range of acceptable handrail heights is derived in the manner described earlier. These results are listed in Table 4.5. For parameter P2, all of the tested heights appear to be satisfactory. The same is true of parameter P4, in the elderly subjects. In all other cases, analyses of parameters P1, P4 and P5 all point to lower limits on handrail height. This is to be expected, since the earlier analysis found that these parameters increase with increasing handrail height. Conversely, analysis of parameter P3 points to an upper limit on handrail height. Again, this is in agreement with the earlier results, which found that P3 decreases with increasing handrail height.

The overall range is defined as a range of handrail heights that falls within the various ranges specified through analysis of the individual parameters. For the young subjects, the overall range includes heights between 36" and 38". For the elderly subjects, the only heights which satisfy all requirements are 36" and 40". However, the discontinuity at 38" is probably not significant (see Figure 4.18); therefore, the overall range for the elderly subjects could probably be modified to include all heights between 36" and 40". Thus, for the general user population, the estimate for the optimal range of handrail heights defines 36" as the lower limit and 38" as the upper limit.

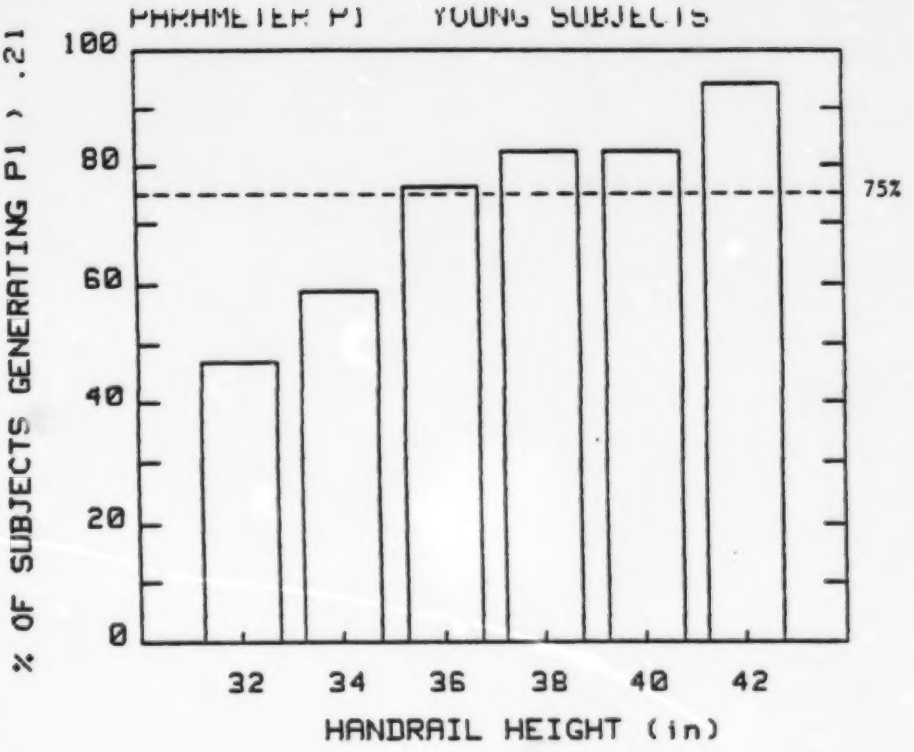
(5) Subjective User Preference

Each subject was asked to select his/her preferred handrail height, based purely on consideration of comfort. The distributions of the responses are plotted for the young and the elderly subject groups in Figure 4.21.

For both young and elderly subjects, the mean preferred handrail height was approximately 36 inches.

TABLE 4.5 - Summary of Design Recommendations

PARAMETER	ACCEPTABLE RANGE OF HANDRAIL HEIGHTS	
	Young Subjects	Elderly Subjects
P1 - Backward Moment	height $\geq$ 36"	height $\geq$ 36"
P2 - Backward Force	height $\geq$ 34"	any height tested
P3 - Upward Force	height $\leq$ 38"	height $\leq$ 36" or height = 40"
P4 - Forward Moment	height $\geq$ 36"	height $\geq$ 34"
P5 - Forward Force	height $\geq$ 36"	any height tested
OVERALL	36" $\leq$ height $\leq$ 38"	height = 36" or height = 40"



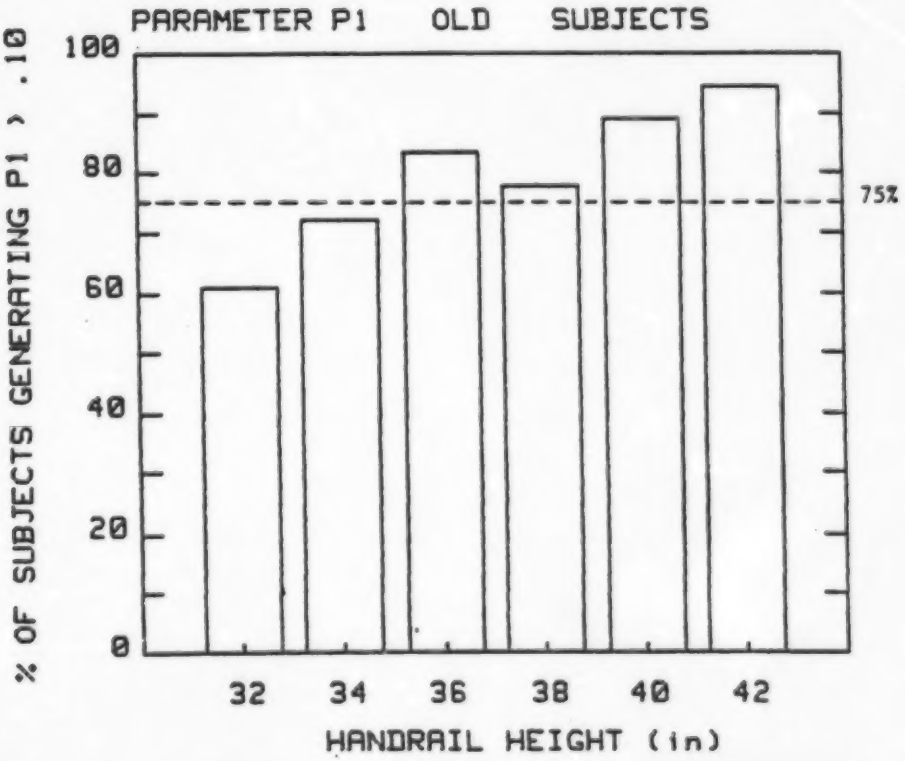


Figure 4.16 Design Range Inferred from Parameter P1 (backward moment)

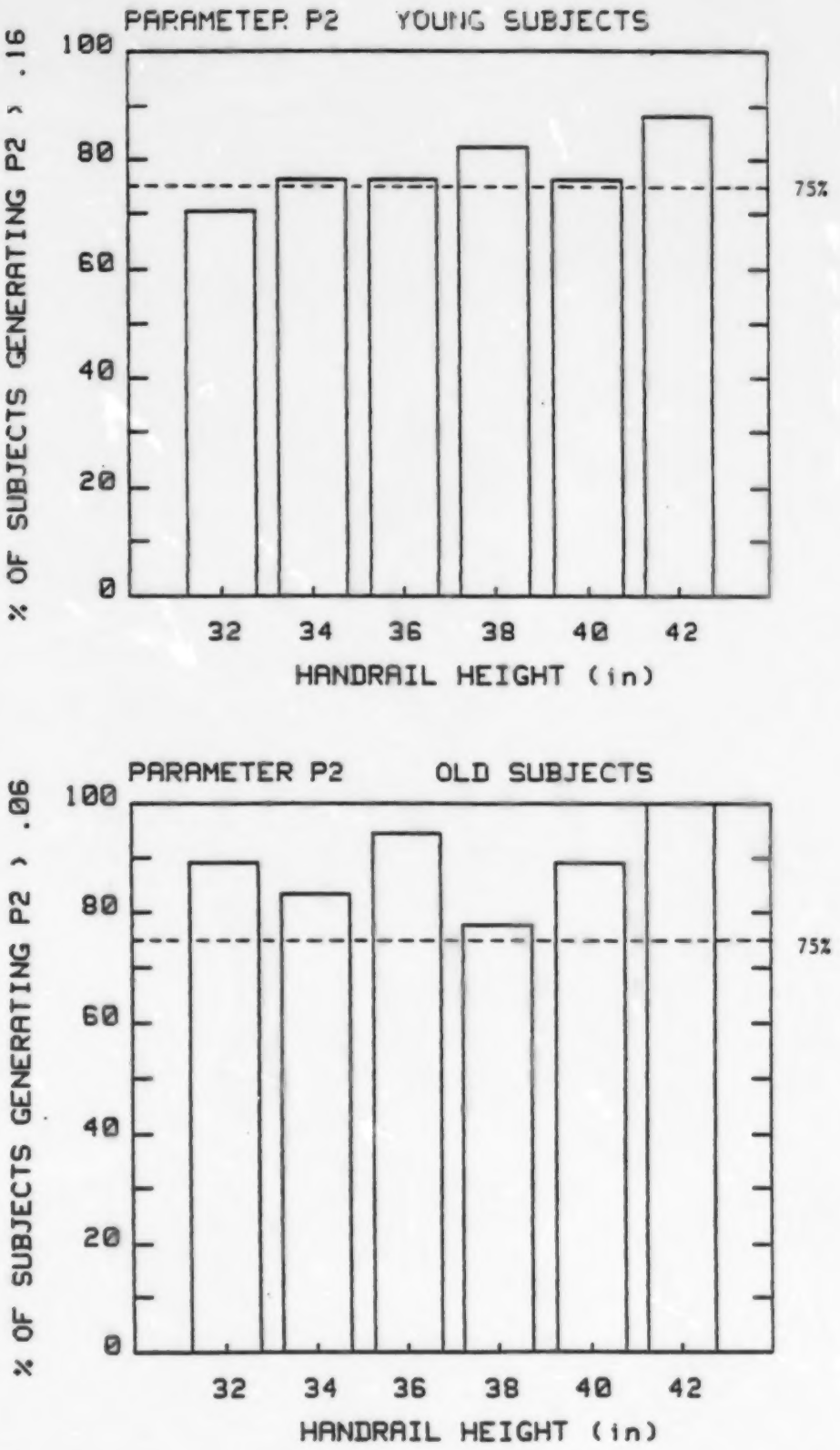
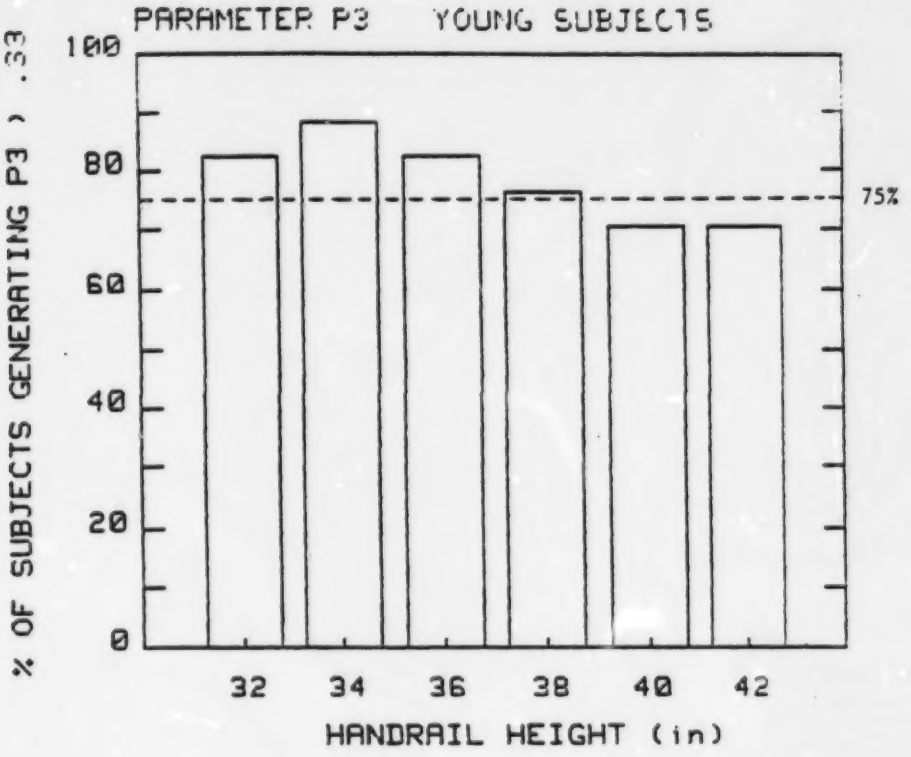


Figure 4.17 Design Range Inferred from Parameter P2 (backward force)



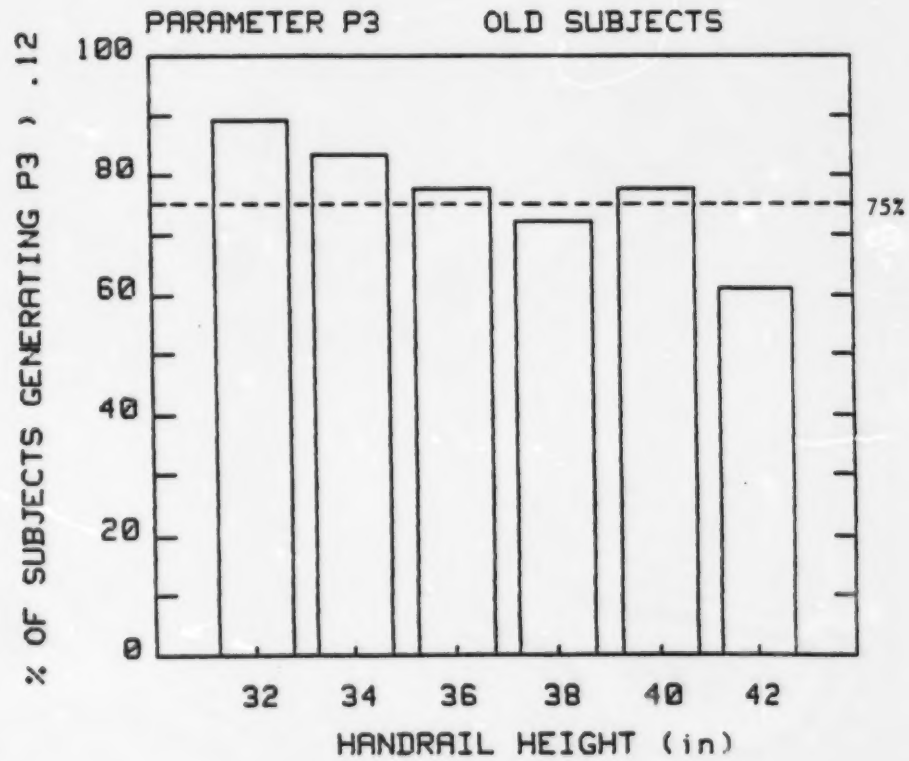
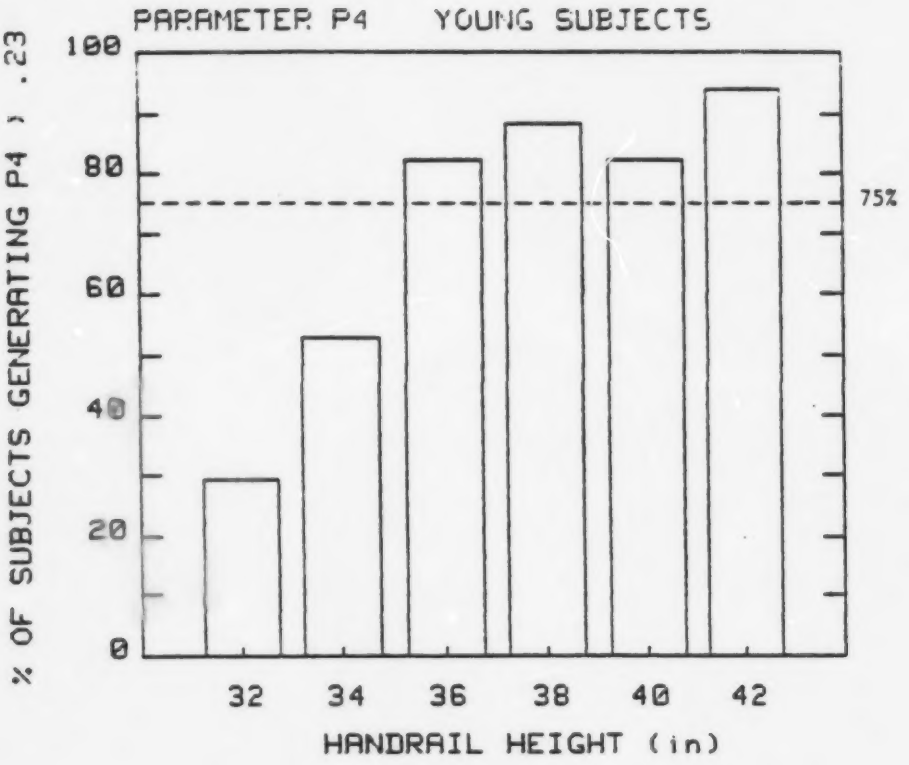


Figure 4.18 Design Range Inferred from Parameter P3 (upward force)



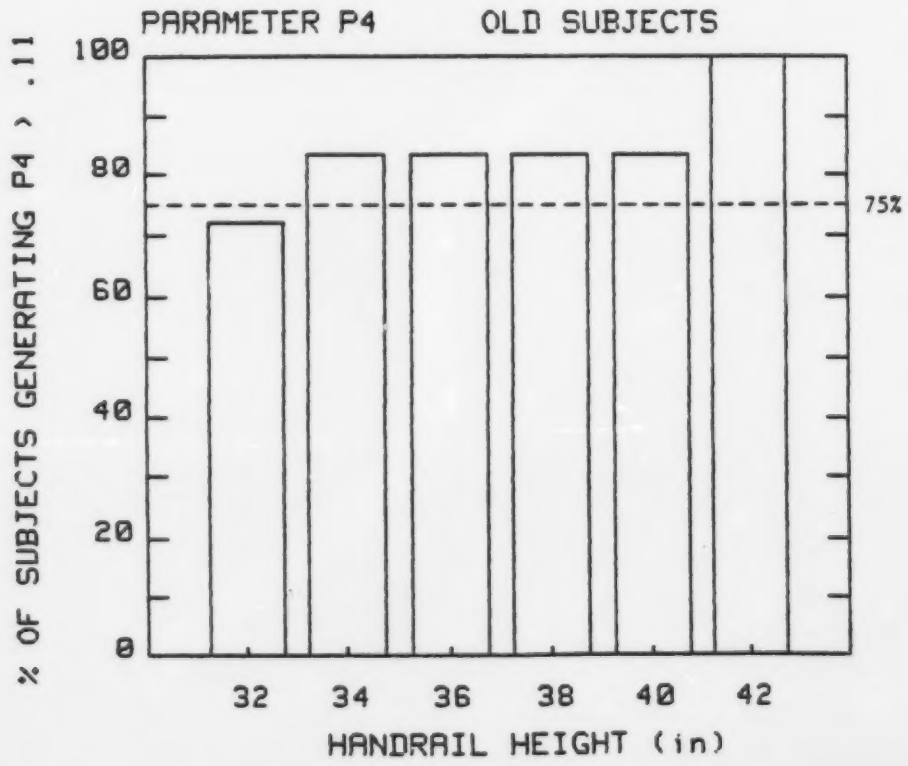


Figure 4.19 Design Range Inferred from Parameter P4 (forward moment)

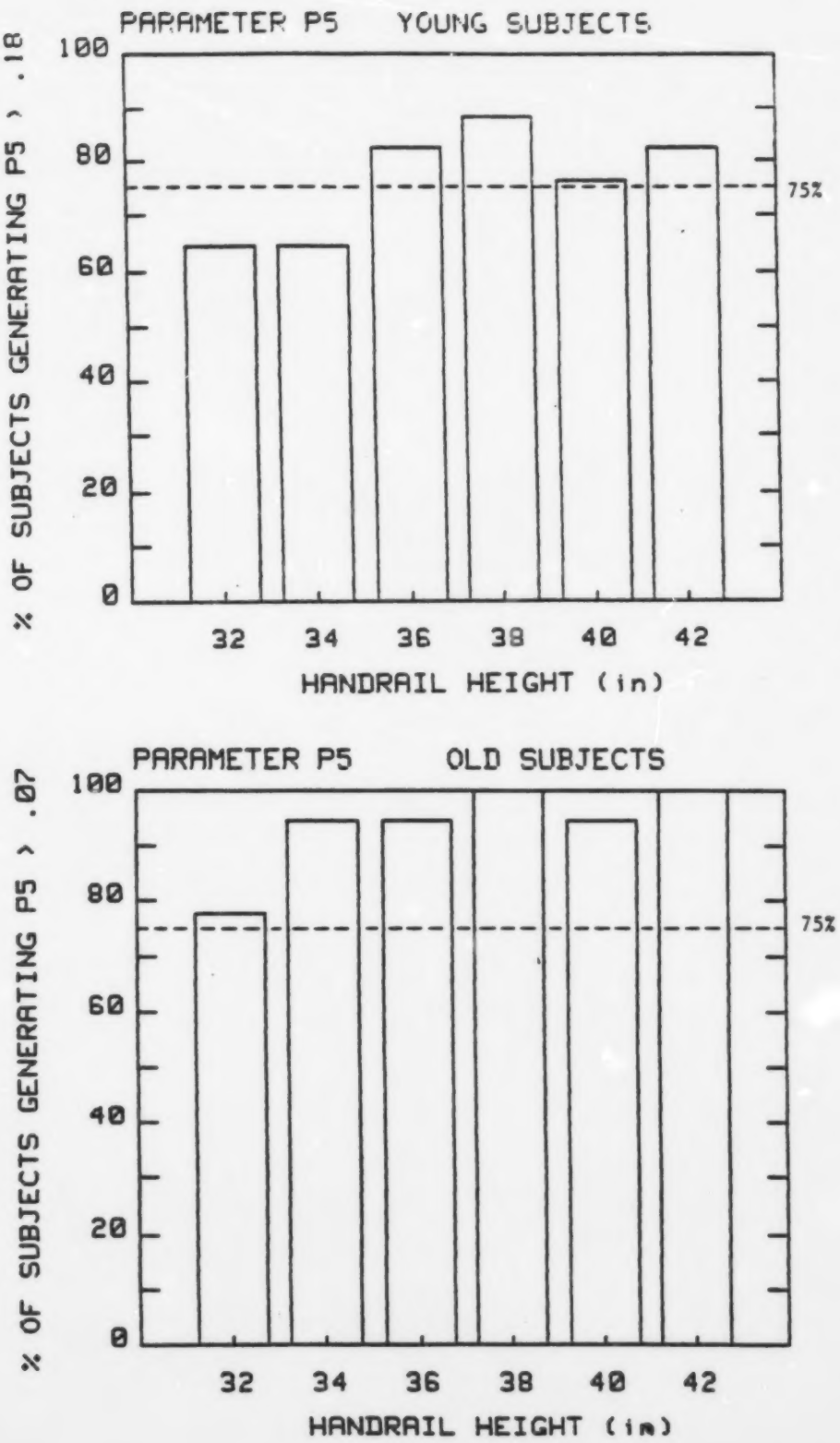


Figure 4.20 Design Range Inferred from Parameter P5 (forward force)

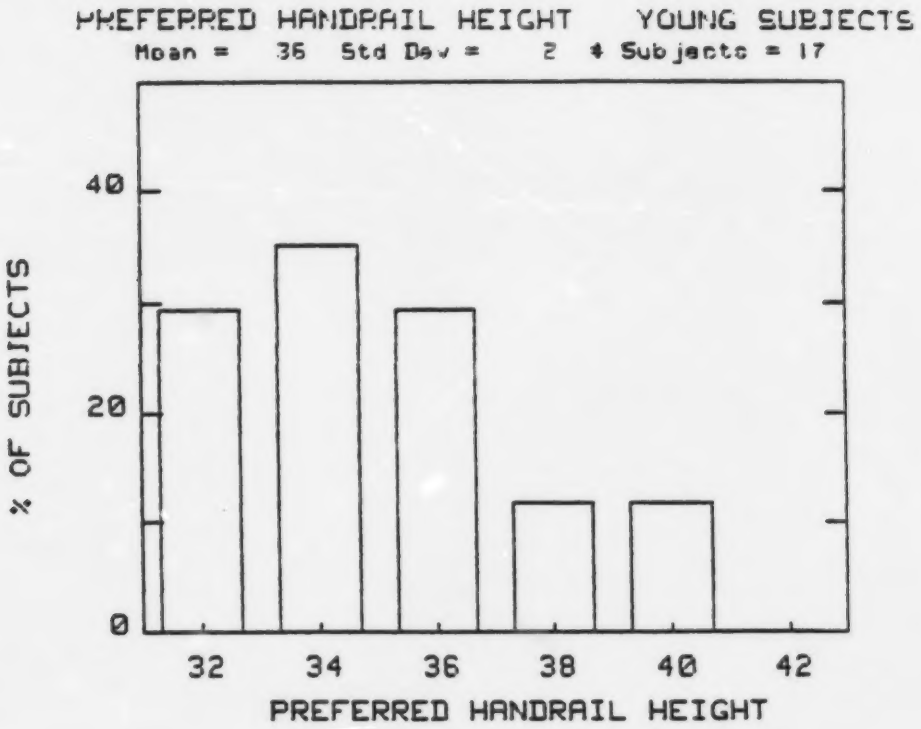


Figure 4.21 (a) Preferred Handrail Height - Young Subjects

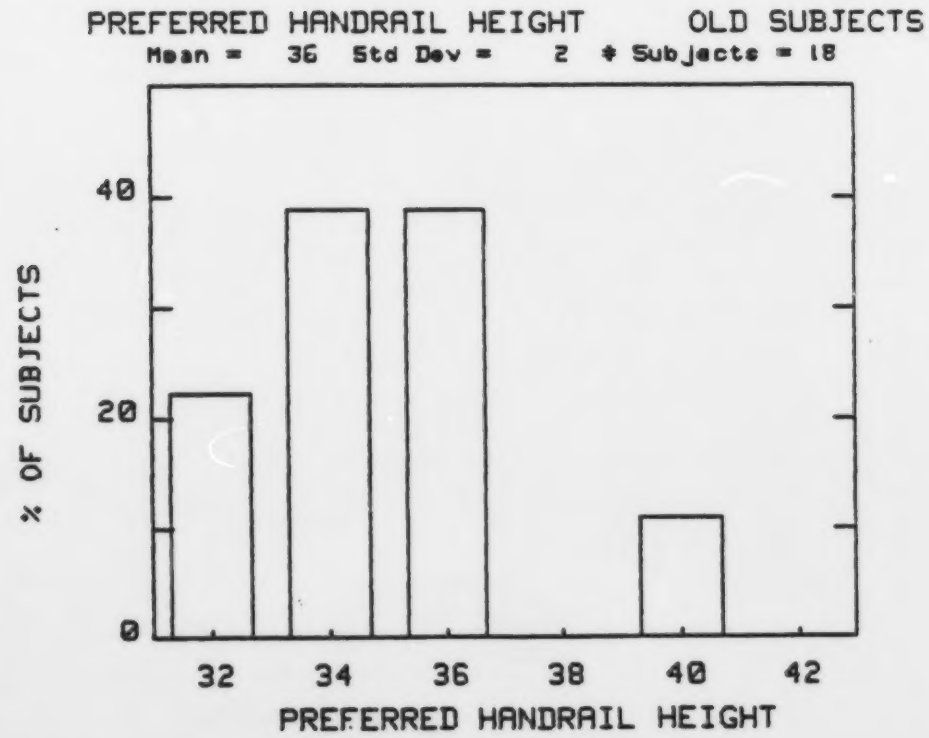


Figure 4.21 (b) Preferred Handrail Height - Elderly Subjects

Discussion

(1) Effects of Handrail Height

Although the statistical analysis demonstrated a significant linear dependence of force/moment generation on handrail height, the magnitude of the dependence must also be considered. Typically, changes in force/moment generation were in the order of 5 to 10% of body weight/body weight moment per 10-inch change in handrail height. The actual stabilizing effect of such changes in force/moment can only be determined through fall simulation experiments, incidence-observation studies, or complex biomechanical modelling.

Measurements of grip location indicated that, almost without exception, subjects tended to move their grip forward as the handrail height was increased. Due to the redundant degrees of freedom of the arm, grip location is not constrained; therefore, some cognitive process must be involved.

Choice of grip location could be based on the individual's past experience with stairway handrails. Based on this experience, the individual may have learned that a forward grip location is more effective in stabilizing his/her body. It would seem likely that the subject's choice of grip location is also based on considerations of comfort of the arm. Presumably, comfort is maximized over some range of wrist, elbow and shoulder angles. Outside this range, discomfort or pain may be experienced. In an attempt to evaluate this hypothesis, estimates of elbow and shoulder flexion angle were made, using measurements of grip location and measurements of body dimensions and position.

The results showed that elbow flexion generally increases only slightly as handrail height is increased. This result would tend to indicate that the individual copes with increases in handrail height by moving his/her grip forward rather than increasing flexion at the elbow.

Moving the grip forward apparently: (1) increases the magnitude of horizontal force generation, presumably as a result of increases in the mechanical advantage of the arm, and (2) increases the moment arm of the vertical force about the metatarsophalangeal joint. Both of these effects act to increase the stabilizing moment about this joint. Vertical force generation decreases as the grip is moved forward, presumably because of decreases in the mechanical advantage of the arm. The decrease in vertical force tends to reduce the stabilizing moments about the metatarsophalangeal joints.

From the foregoing discussion, the advantages of the experimental approach should be evident. First of all, biomechanical modelling would not be able to predict grip location, since the mechanical system is redundant. Secondly, even if grip location were known, it would be very difficult to predict the overall results, due to the conflicting effects of changes in moment arms, muscle lengths and the mechanical advantage of the arm and hand segments. A very sophisticated model would be required in order to attain accurate results.

(2) Age Differences

The significant age differences discussed in the previous section could be due to a number of factors. Lower levels of force and moment can be explained in terms of the well-known decreases in muscle strength with age. Lower levels of non-dimensionalized force may be due to this effect, or to the fact that the elderly individuals are more likely to be overweight.

Age-related reduction in stature might be expected to result in the elderly subjects choosing more forward grip locations; reduction in range of elbow rotation might also have the same effect. In either case, a more forward grip location seems to increase the horizontal force, decrease the vertical force and increase the moment about the metatarsophalangeal joints, as discussed earlier. Apparently, any increases created by the more advantageous grip location are outweighed by the decreases in muscular strength, on the average.

The differences in grip location may also explain the differences between the two age groups in the dependence of their force/moment generating capability on handrail height. Moment generation appears to be particularily sensitive to grip location. Greater variation in elbow flexibility in the elderly subjects probably results in greater variability in grip location, and may thereby explain why the elderly subjects fail to demonstrate a common slope in the dependence of moment generation on handrail height.

Some of the age differences could be an artifact of the experimental protocol. For example, fatigue or learning effects could be more pronounced in the elderly subjects. One might also expect elderly subjects to experience greater motivational variation. Confusion regarding the instructions and purpose of the experiment might have a similar effect (some confusion was evident in some of the elderly subjects).

Another factor which might explain some of the observed age differences would be the differences in the relative numbers of males and females in each group. (The male-to-female ratio was 2.4 in the young group, and 1.25 in the elderly group.) Males are generally able to generate larger forces, although non-dimensionalization with respect to body size should compensate for this effect to some extent.

(3) Estimation of the Optimal Design Range

The method by which the optimal design range was calculated does not lend itself to a statistical analysis of significance. Furthermore, in calculating the design range, the definition of "acceptability" (based on 75% of the subjects) was chosen solely to allow a reasonably narrow range to be specified. Therefore, the results should be treated as an estimate of the optimal design range for handrail height. The results can be extrapolated to the general user population only to the extent that the subject population is representative of the general population, and to the extent that motivational variation (error) can be neglected.

However, there is reason to have confidence in the results. The earlier statistical analysis demonstrated a significant linear relationship between each of the five force and moment parameters and handrail height. The histograms used in estimating the optimal range tend to follow the same trends. Thus, histograms for parameters P1, P2, P4, and P5 tend to increase with increasing handrail height, while the histograms for P3 tend to decrease with increasing handrail height.

Consideration of subjective user preference is important. If the stairway users feel that a particular handrail is uncomfortable or awkward to use, then they may decide not to use the handrail. In this situation, it is much more difficult to recover from a loss of balance without falling, due to the time delay associated with the grabbing response. Thus, the biomechanical advantages of a particular handrail could be negated.

Unfortunately, it was not possible to establish a design range based on user preference. Many subjects had difficulty in identifying their preferred height, much less specifying an acceptable range. Thus, the analysis of user preference yielded only the mean value of preferred height for each age group. This is not a suitable basis for design, since the mean could be unacceptable for a very large percentage of the user population. However, it is encouraging that both means fall within the design range recommended on the basis of the biomechanical measurements.

(4) Limitations of the Present Study

(i) Variability in the Results

Although the statistical analysis yielded significant results, the variability in the results was too large to allow a statistical prediction of optimal handrail height or to perform correlations with body size and weight. Reduction in the variability might allow these forms of analysis to be performed.

"Within-subject variability is due to learning effects, fatigue, changes in the tightness of the restraining harness, and psychological factors occurring during the experimental session. "Between-subject" variability is due to differences between subjects in any of the above factors.

Learning effects were found to be significant, but only within the first one or two trials. Therefore, by re-testing the initial handrail height and treating the initial test as a learning trials, variation due to learning was minimized. Fatigue was minimized by limiting the duration of the test (30 to 35 minutes) and by providing frequent rest periods during the test. These strategies were apparently successful, as no subjects complained of fatigue and no subjects elected to take advantage of the offer to take a sitting rest period midway through the experimental session. Both fatigue and

learning effects were also minimized by randomizing the order at which the different handrail heights were tested. By doing this, variation due to fatigue and learning effects will "cancel out" in the calculation of mean values.

Loosening of the restraining harness was controlled by periodic checks during the experimental session; however, differences between subjects in the tightness of the restraining harness is difficult to control, and may be one major cause of variability. One subject in particular, was able to use his body weight to augment the force on the handrail. The effect was obvious in this case, and this subject was eliminated from the analysis of covariance.

One important pyschological factor is the selection of grip location. By conducting the tests on a mock stairway and by instructing the subjects to imagine that they were descending the stairway, we endeavored to provide a sufficiently realistic simulation of actual stairway use, so that the subjects could make realistic selections of grip position.

Variation in psychological motivation (in exerting maximal force) is a major problem. Obvious factors, such as the instructions given to the subjects, encouragement from the experimentor, and the presence of spectators were all strictly controlled. Otherwise, motivational error cannot be controlled.

Given certain unavoidable sources of error, the reliability of the experimental results can still be improved, by either increasing the number of subjects in the study, or by repeating the trials for each subject. Time constraints prohibited us from using either of these approaches.

(ii) Extrapolation of Results

The purpose of the experiments is to draw conclusions that can be extrapolated to actual stairway use by the general population. The question of extrapolation from our subject population to the general population was dealt with by selecting subjects (within each age group) at random. Thus, given a sufficient number of subjects, the characteristics of the subject population should approach the characterstics of the general population. The similarity between the subject and general populations (see Figures 4.2 to 4.4) is evidence that we did have sufficient numbers of subjects. One exception is in the male-to-female ratio, which was abnormally large in our subject population (particularily in the young subject group); however, the fact that all force and moment parameters were non-dimensionalized with respect to body weight and height should minimize this problem.

Extrapolation of the results to actual stairway use is another matter. As discussed in Section III, the erect posture is a reasonable approximation of body position at the beginning of a fall.

Thus, the static force measurements made in this position should give an indication of the maximum stabilizing forces and moments that could be generated at the start of a fall. Measurement of maximal values provides a suitable basis for comparing different handrail designs; however, there is no implication that these levels of force and moment would actually be generated during a fall. Other factors (e.g. time delay in the individual's response, psychological factors) may prevent the subject from attaining his/her maximal values.

Furthermore, as the body proceeds to fall forward, the actual forces and moments that can be exerted will change due to positional changes of the body relative to the handrail. In addition, muscle force generation will change under dynamic conditions (the present tests were essentially isometric measurements). Nonetheless, it is anticipated that the trends (in comparing the different handrail heights) indicated by the static, erect measurements may hold for other stages of the fall. Even if this is not the case, generation of stabilizing forces and moments is probably most crucial in the initial stages of the fall, when the body is still relatively erect and dynamic effects are small--conditions replicated in out experimental protocol.

Conclusions

(1) Handrail height has a significant influence on the ability to generate stabilizing forces and moments.

(2) The dependence of this ability to generate stabilizing forces and moments varies linearly with handrail height.

(i) For backward and forward moment and for backward and forward force, the non-dimensionalized magnitude of the force or moment increases with increasing handrail height.

(ii) For upward force, the non-dimensionalized magnitude decreases with increasing handrail height.

(iii) For each of the non-dimensionalized forces, all individuals have the same linear dependence on handrail height (i.e a common slope at each parameter).

(iv) For each of the non-dimensionalized forces and moments, all young individuals have the same linear dependence on handrail height (i.e a common slope at each parameter).

(v) The "relative magnitudes" (i.e. y-intercepts) of the forces and moments generated by each individual are significantly different from those generated by other individuals.

(3) There are signficant age-related differences in the ability to generate stabilizing forces and moments.

(i) On the average, young individuals exert larger forces and moments, relative to body weight and size, than elderly individuals.

(ii) In generating stabilizing moments (backward and forward), young individuals demonstrate a common linear dependence on handrail height. In generating stabilizing moments, elderly individuals vary significantly in their dependence on handrail height.

(iii) In generating stabilizing forces (forward, backward and upward), both young and elderly individuals demonstrate a common linear dependence on handrail height.

(4) The optimal design range for handrail heights is estimated to be substantially the same for both young and elderly subjects.

(i) The lower limit on handrail height is estimated to be 36 inches (0.914 meters).

(ii) The upper limit on handrail height is estimated to be 38 inches (0.965 meters).

(5) In terms of comfort, the mean preferred handrail height of both the young and elderly populations is approximately 36 inches (0.914 meters).

Note: All of the above conclusions were derived from experiments conducted at the subject's preferred lateral displacement, on a cylindrical oak handrail (1.75" in diameter) at a slope of 33 degrees, as described in the protocol. Variation in lateral displacement or handrail shape, size, material, or slope could change the results.

V DIRECTIONS FOR FUTURE RESEARCH

There are at least four directions in which future research could proceed: (1) verification of the present results, (2) investigation of the effects of other handrail parameters, using the present experimental paradigm, (3) investigation of the effect of handrail parameters on the grabbing response, and (4) investigation of the effects of handrail parameters on incidence of missteps and slips.

Verification of the Present Results

The applicability of the present results to actual fall-prevention could conceivably be verified through field or laboratory studies of the incidence of falling. This approach was discussed in general in Section III, and the disadvantages noted there still apply, namely, the need for large numbers of subjects and/or large numbers of observed trials. However, there is no longer a need to investigate a wide range of handrail heights. Instead, an optimal height recommended here (36 to 38 inches) could be compared with one or two other values.

Investigation of Other Handrail Parameters

There is no reason why the present experimental paradigm could not be used to investigate a number of other handrail parameters, such as handrail shape, size, compliance, and surface texture. The present protocol would be used to compare various alternatives. Note that start-up costs would be minimal, as the present experimental apparatus was designed to allow these types of parameters to be studied and could be used without any further modification.

All of the afore-mentioned parameters are felt to be of prime importance in the generation of stabilizing forces and moments. However, compliance of the handrail material and texture of the handrail surface are two factors which do not seem to have received any serious attention in the literature. By combining the present force and moment measurements with tests of user preference, it is likely that significant and useful results could be achieved at minimal cost.

In addition, it is necessary to determine the optimal handrail height at different stairway slopes, since changes in slope may well affect the optimal value. Of particular importance would be slopes in the order of 45 to 49 degrees, which are currently in common use in residential buildings. The experimental results could indicate a need for retro-fitting of these buildings with more suitable handrails.

Investigation of the Grabbing Response

The grabbing response is important in stairway falls where the user has not maintained continuous contact with the handrail. The major issue is felt to be the trade-off between speed and accuracy of the grabbing response. The handrail design parameters which are most likely to affect this trade-off are wall clearance, spacing between parallel handrails, handrail size and handrail height.

An experimental investigation could study the speed-accuracy trade-off, by measuring the speed and accuracy of the grabbing response. Accuracy could be assessed by recording the incidence of misgrabs and by measuring the force/moment generation achieved in cases where the handrail is grabbed successfully. The trials could be initiated by means of a visual or auditory stimulus, or alternatively, by provoking a loss of balance through the application of a postural perturbation to the subject. The fall simulation approach would be more realistic; however, due to safety problems discussed in Section III, this approach would not be suitable for elderly subjects.

Of particular importance would be determination of minimal wall clearance and maximal handrail spacing. In addition to safety considerations, both factors play a major role in determining traffic flow rates and in determining building costs. Minimization of wall clearance requirements and maximization of spacing requirements would significantly reduce building costs.

Investigation of Missteps and Slips

The present study has focussed on the problem of fall recovery, that is, stabilization of the body after an initial loss of balance has occurred. The other issue, which has not been addressed to any significant degree in the literature, is the effect of handrail parameters on the incidence of the initial loss of balance (i.e. slips and missteps).

As mentioned in Section III, the effect of handrail parameters on the probability of slipping could be assessed through the use of force plate measurements. A reduction in anterior-posterior shearing force while traversing the test stairway would be indicative of a reduced probability of slipping. Unfortunately, missteps could only be studied through an incidence-observation approach, conducted in either a laboratory or field setting. As discussed earlier, the chief disadvantages of this approach are the large numbers of subjects and/or trials required. Visual or auditory disturbances could be used to accelerate the incidence of missteps and/or slips; however, this could result in serious safety concerns, particularily for elderly subjects.

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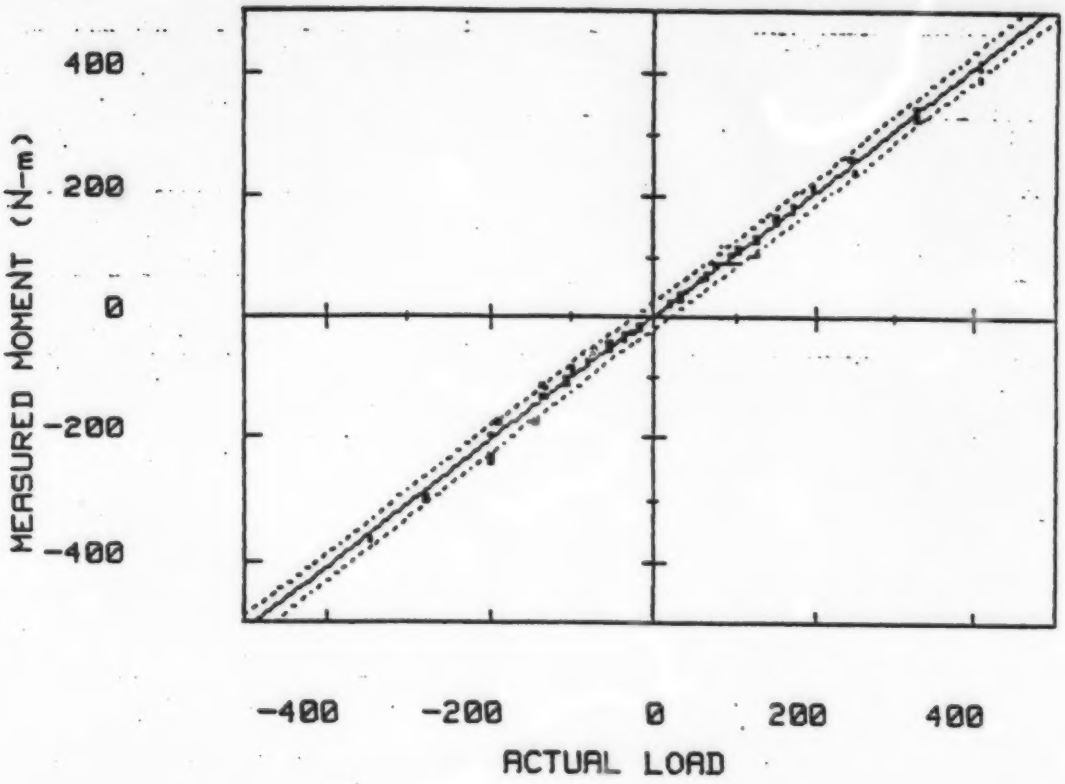
APPENDIX A

CALIBRATION OF THE FORCE PLATE

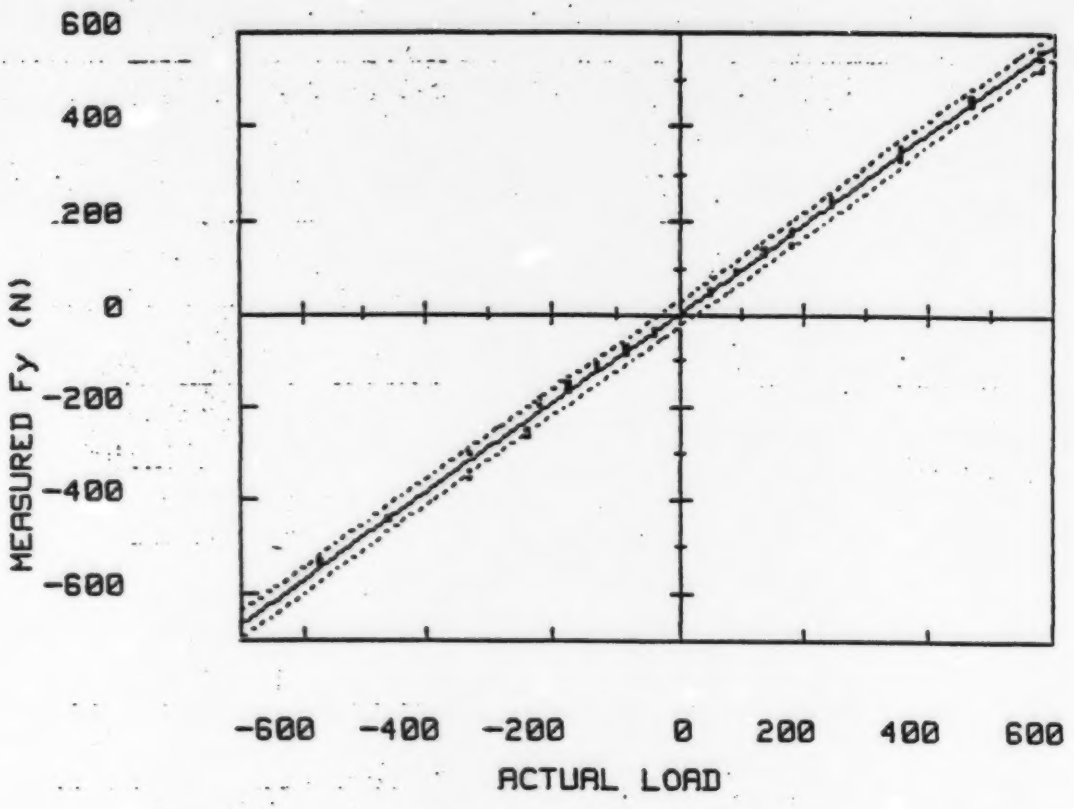
CALIBRATION OF MOMENT



OFFSET = 2.03494; SLOPE = 1.02939
CORRELATION COEFFICIENT = .99755

MEAN ERROR = 8.4%; MAX ERROR = 24.0%

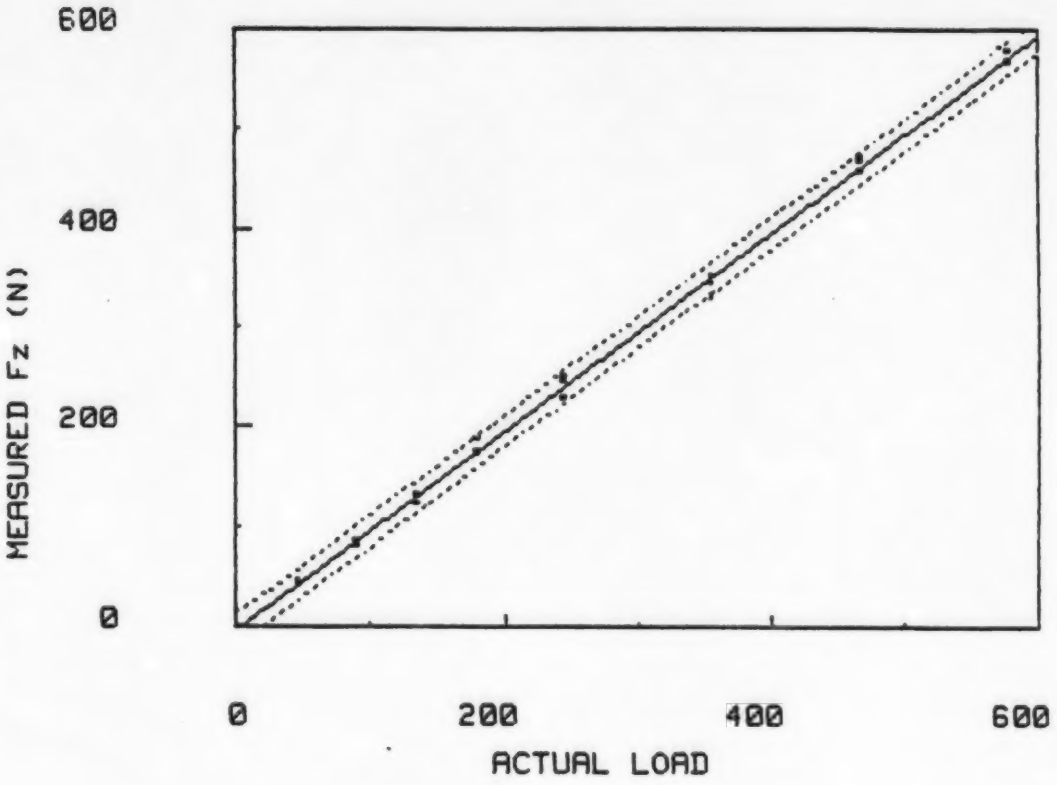
CALIBRATION OF HORIZONTAL FORCE



OFFSET = 1.75161; SLOPE = .95612
CORRELATION COEFFICIENT = .99879

MEAN ERROR = 6.8%; MAX ERROR = 21.2%

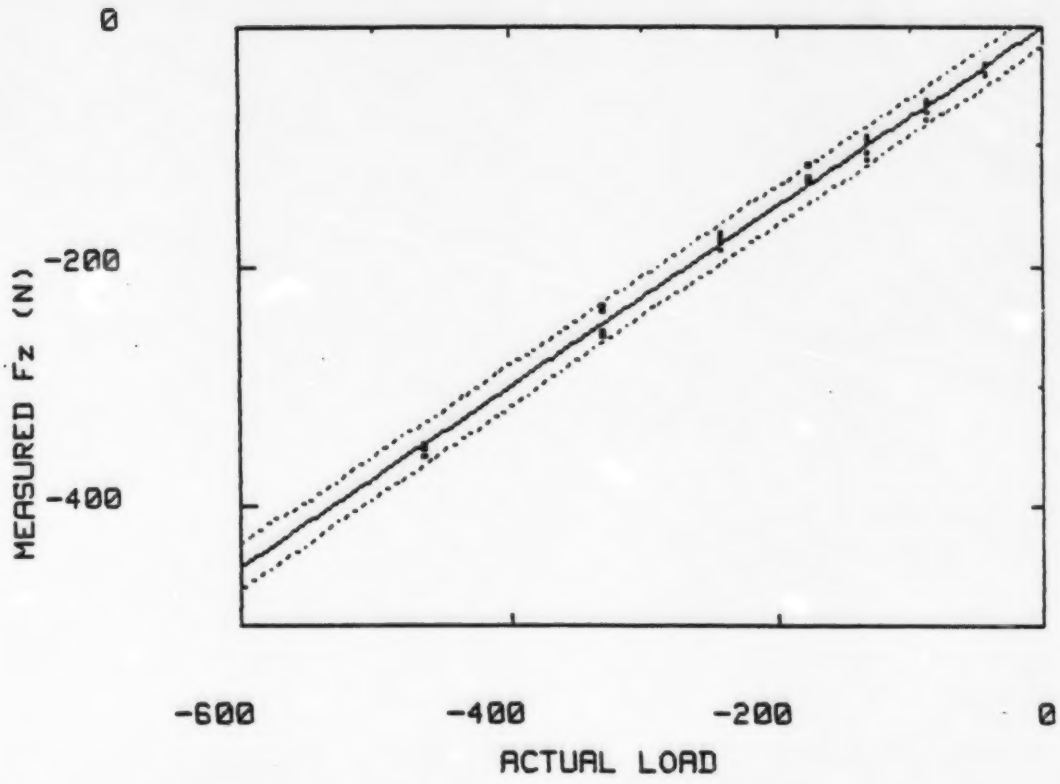
CALIBRATION OF VERTICAL FORCE



OFFSET = -3.22968; SLOPE = .99320
CORRELATION COEFFICIENT = .99908

MEAN ERROR = 4.7%; MAX ERROR = 18.1%

CALIBRATION OF VERTICAL FORCE

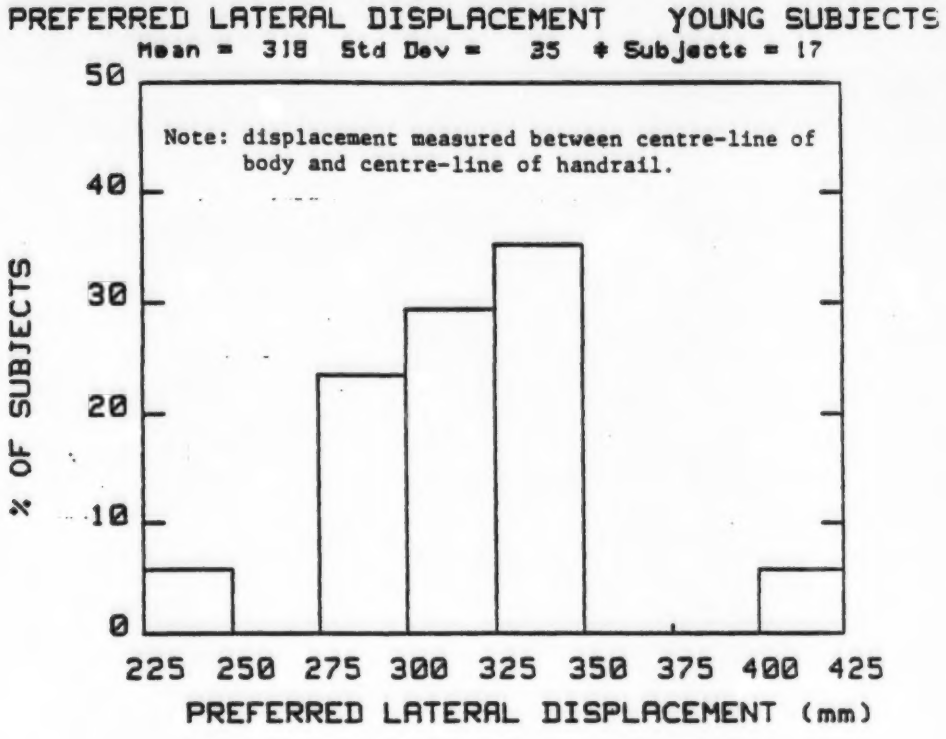


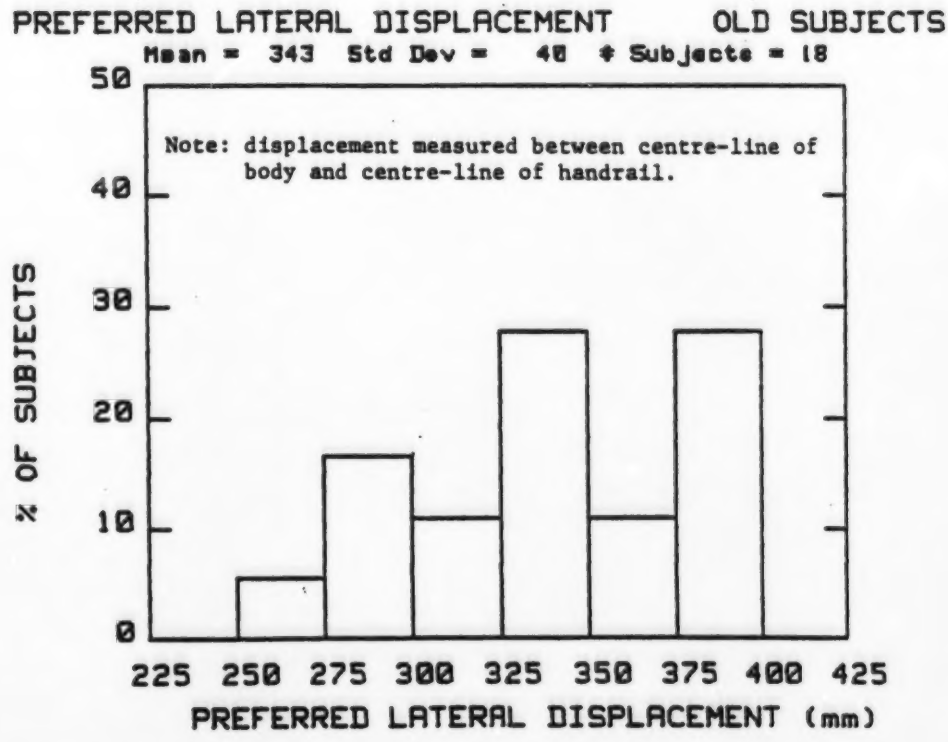
OFFSET = -.23047; SLOPE = .74984
CORRELATION COEFFICIENT = .99719

MEAN ERROR = 23.9%; MAX ERROR = 34.2%

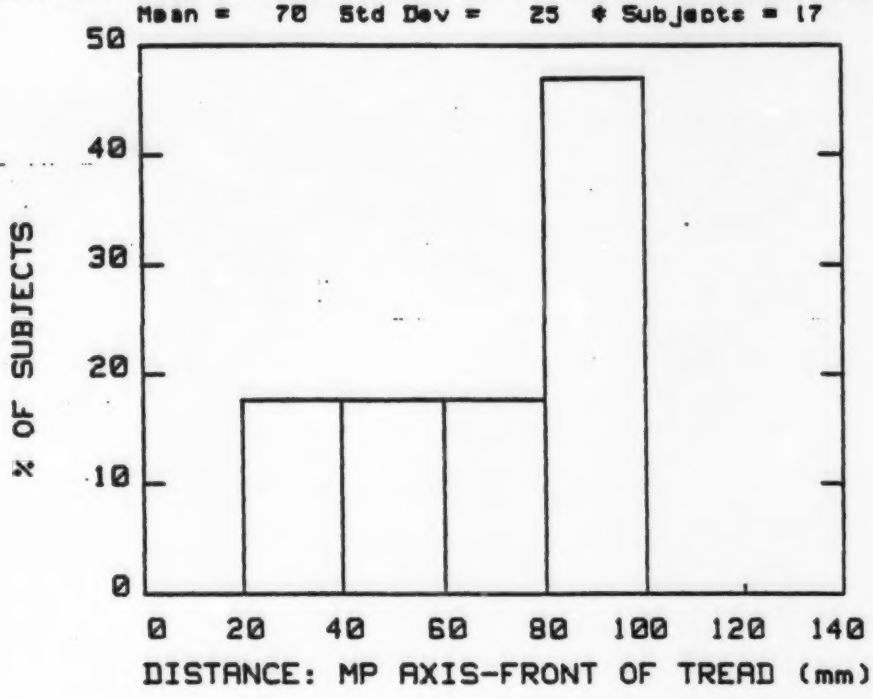
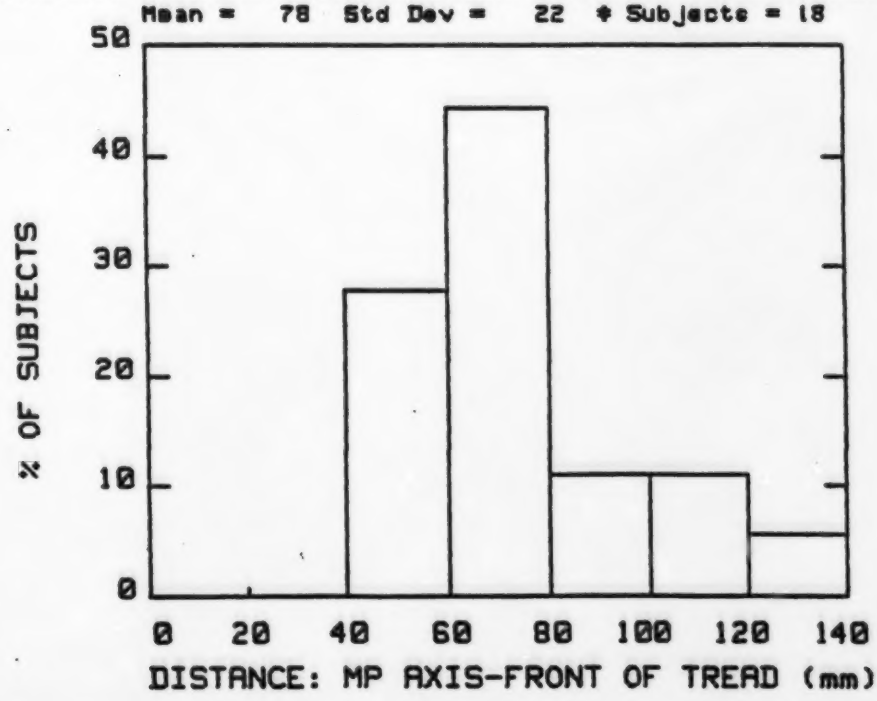
APPENDIX B

PREFERRED BODY POSITION


PREFERRED LATERAL DISPLACEMENT YOUNG SUBJECTS
Mean = 318 Std Dev = 35 # Subjects = 17
Note: displacement measured between centre-line of body and centre-line of handrail.
% OF SUBJECTS
0
10
20
30
40
50
225 250 275 300 325 350 375 400 425
PREFERRED LATERAL DISPLACEMENT (mm)


PREFERRED LATERAL DISPLACEMENT OLD SUBJECTS
Mean = 343 Std Dev = 40 # Subjects = 18
Note: displacement measured between centre-line of body and centre-line of handrail.
% OF SUBJECTS
0
10
20
30
40
50
225 250 275 300 325 350 375 400 425
PREFERRED LATERAL DISPLACEMENT (mm)

LOCATION: METATARSOPHALANGEAL AXIS . YOUNG SUBJECTS



LOCATION: METATARSOPHALANGEAL AXIS OLD SUBJECTS